3

BASIC PRINCIPLES OF VENTILATOR DESIGN

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THE VENTILATOR AS A "BLACK BOX"

A mechanical ventilator is an automatic machine designed to provide all or part of the work the body must do to move gas into and out of the lungs. The act of moving air into and out of the lungs is called breathing, or, more formally, ventilation.

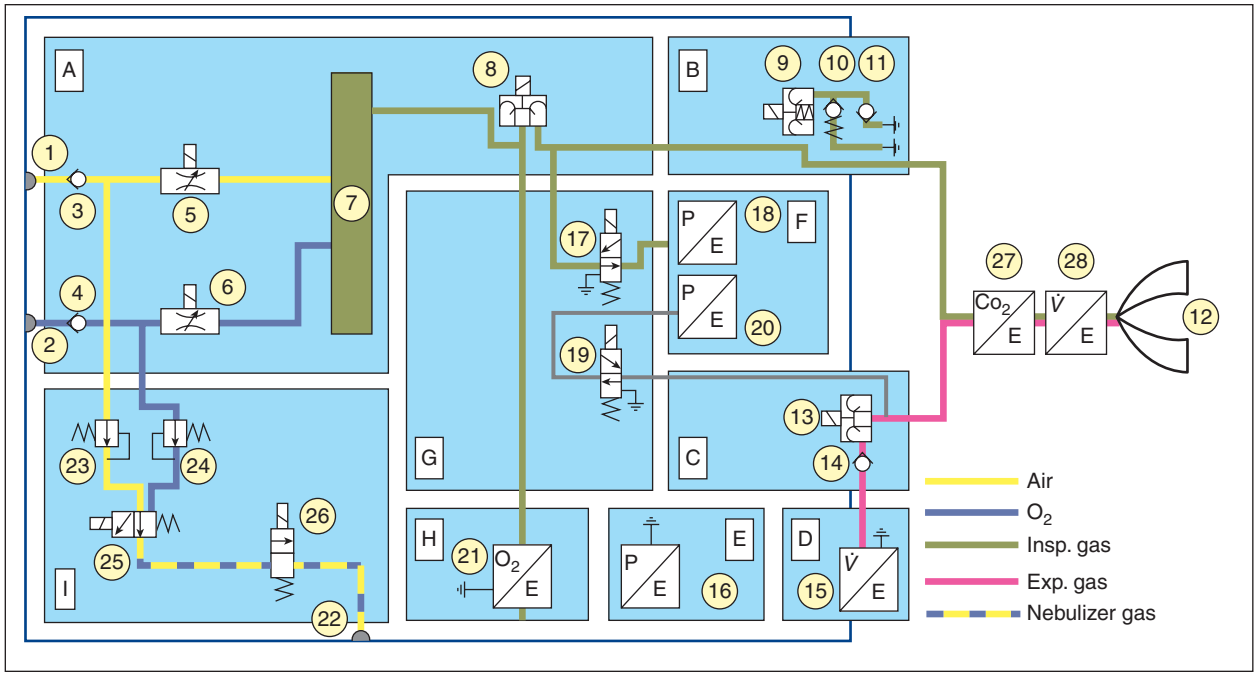
The simplest mechanical device we could devise to assist a person's breathing would be a hand-driven, syringe-type pump that is fitted to the person's mouth and nose using a mask. A variation of this is the self-inflating, elastic resuscitation bag. Both of these require one-way valve arrangements to cause air to flow from the device into the lungs when the device is compressed, and out from the lungs to the atmosphere as the device is expanded. These arrangements are not automatic, requiring an operator to supply the energy to push the gas into the lungs through the mouth and nose. Thus, such devices are not considered mechanical ventilators.

Automating the ventilator so that continual operator intervention is not needed for safe, desired operation requires three basic components:

1. A source of input energy to drive the device;
2. A means of converting input energy into output energy in the form of pressure and flow to regulate the timing and size of breaths; and
3. A means of monitoring the output performance of the device and the condition of the patient.

There was a time when you could take a handful of simple tools and do routine maintenance on your car engine. About that time the average clinician could also completely disassemble and reassemble a mechanical ventilator as a training exercise or to perform repairs. In those days (the late 1970s), textbooks[1] describing ventilators understandably paid much attention to the individual mechanical components and pneumatic schematics. In fact, this philosophy was reflected to some extent in previous editions of this book. Today, both cars and ventilators are incredibly complex mechanical devices controlled by multiple microprocessors running sophisticated software (Fig. 3-1). Figure 3-2 shows the pneumatic schematic of a current intensive care ventilator. All but the most rudimentary maintenance of ventilators is now the responsibility of specially trained biomedical engineers. Our approach to describing ventilator design has thus changed from a focus on individual components to a more generalized model of a ventilator as a "black box," that is, a device for which we supply an input and expect a certain output and whose internal operations are largely unknowable, indeed, irrelevant, to most clinical operators. What follows, then, is only a brief overview of the key design features of mechanical ventilators with an emphasis on input power requirements, transfer functions (pneumatic and electronic control systems), and outputs (pressure, volume, and flow waveforms). The rest of the chapter focuses on the interactions between the operator and the ventilator

FIGURE 3-1 Examples of commonly used intensive care ventilators: **A.** Dräger Infinity V500, **B.** Hamilton G5, **C.** Maquet Servo i, **D.** Covidien PB840. (Image with permission from Nellcor Puritan Bennett LLC, Boulder, Colorado, doing business with Covidien.)



1 Air gas inlet
2 O_2 gas inlet
3 Air nonreturn valve
4 O_2 nonreturn valve
5 Air metering valve
6 O_2 metering valve
7 Tank
8 Mixed gas metering valve
9 Safety valve
10 Emergency expiratory valve
11 Emergency breathing valve
12 Patient's lungs
13 Expiratory valve
14 Nonreturn valve
15 Expiratory flow sensor
16 Barometric pressure sensor
17 Calibration valve for inspiratory pressure sensor
18 Inspiratory pressure sensor
19 Calibration valve for expiratory pressure sensor
20 Expiratory pressure sensor
21 O_2 sensor
22 Nebulizer outlet
23 Air pressure regulator
24 O_2 pressure regulator
25 Nebulizer mixer valve
26 Nebulizer changeover valve
27 CO_2 sensor
28 Neonatal flow sensor (depending on the patient category)

FIGURE 3-2 Pneumatic schematic of the Dräger Infinity V500 intensive care ventilator. **A.** Gas-mixture and gas-metering assembly. Gas from the supply lines enters the ventilator via the gas-inlet connections for oxygen and air (*1,2*). Two nonreturn valves (*3,4*) prevent one gas from returning to the supply line of the other gas. Mixing takes place in the tank (*7*) and is controlled by two valves (*5,6*). Inspiratory flow is controlled by a third valve (*8*). **B.** Inspiratory unit consists of safety valve (*9*) and two nonreturn valves (*10,11*). In normal operation, the safety valve is closed so that inspiratory flow is supplied to the patient's lungs (*12*). During standby, the safety valve is open and enables spontaneous inspiration by the emergency breathing valve (*11*). The emergency expiratory valve (*10*) provides a second channel for expiration when the expiratory valve (*13*) is blocked. **C.** Expiratory unit consists of the expiratory valve (*13*) and a nonreturn valve (*14*). The expiratory valve is a proportional valve and is used to adjust the pressure in the patient circuit. In conjunction with the spring-loaded valve of the emergency air outlet (*10*), the nonreturn valve (*14*) prevents pendulum breathing during spontaneous breathing. **D.** Expiratory flow sensor. **E.** Barometric pressure sensor. Conversion of mass flow to volume, body temperature and pressure saturated (*BTPS*) requires knowledge of ambient pressure. **F.** Pressure measurement assembly. Pressure in the patient circuit is measured with two independent pressure sensors (*18,20*). **G.** Calibration assembly. The pressure sensors are regularly zero calibrated by connection to ambient pressure via the two calibration valves (*17,19*). **H.** Oxygen sensor. **I.** Medication nebulizer assembly. (Reproduced, with permission, from Dräger Medical AG & Co. KG. *V500 Operator's Manual*. Luebeck, Germany.)

(the operator interface), and between the ventilator and the patient (the patient interface).

Inputs

Mechanical ventilators are typically powered by electricity or compressed gas. Electricity, either from wall outlets (e.g., 100 to 240 volts AC, at 50/60 Hz) or from batteries (e.g., 10 to 30 volts DC), is used to run compressors of various types. Batteries are commonly used as the primary power source in the home-care environment but are usually reserved for patient transport or emergency use in hospitals. These sources provide compressed air for motive power as well as air for breathing. Alternatively, the power to expand the lungs is supplied by compressed gas from tanks, or from wall outlets in the hospital (e.g., 30 to 80 pounds per square inch [psi]). Some transport and emergency ventilators use compressed gas to power both lung inflation and the control circuitry. For these ventilators, knowledge of gas consumption is critical when using cylinders of compressed gas.

The ventilator is generally connected to separate sources of compressed air and compressed oxygen. In the United States, hospital wall outlets supply air and oxygen at 50 psi, although most ventilators have internal regulators to reduce this pressure to a lower level (e.g., 20 psi). This permits the delivery of a range of oxygen concentrations to support the needs of sick patients. Because compressed gas has all moisture removed, the gas delivered to the patient must be warmed and humidified so as to avoid drying out the lung tissue.

Conversion and Control

The input power of a ventilator must be converted to a predefined output of pressure and flow. There are several key systems required for this process. If the only power input is electrical, the ventilator must use a compressor or blower to generate the required pressure and flow. A compressor is a machine for moving a relatively low flow of gas to a storage container at a higher level of pressure (e.g., 20 psi). A blower is a machine for generating relatively larger flows of gas as the direct ventilator output with a relatively moderate increase of pressure (e.g., 2 psi). Compressors are generally found on intensive care ventilators whereas blowers are used on home-care and transport ventilators. Compressors are typically larger and consume more electrical power than blowers, hence the use of the latter on small, portable devices.

FLOW-CONTROL VALVES

To control the flow of gas from a compressor, ventilator engineers use a variety of flow-control valves, from very simple to very complex. The simplest valve is just a fixed orifice flow resistor that permits setting a constant flow to the external tubing that conducts the gas to the patient, called the *patient circuit*. Such devices are used in small transport ventilators and automatic resuscitators. Manually adjusted variable-orifice flow meters have been used in simple infant ventilators in the past (e.g., Bourns BP-200) and are currently used in the Infant Flow SiPAP device (CareFusion, Minneapolis, MN), as shown in Figure 3-3. The advent of inexpensive microprocessors in the 1980s led to development of digital control of flow valves that allow a great deal of flexibility in shaping the ventilator's output pressure, volume, and flow waveforms (Fig 3-4).[2] Such valves are used in most of the current generation of intensive care ventilators.

FIGURE 3-3 CareFusion Infant Flow SiPAP device.

Directing flow from the source gas into the patient requires the coordination of the output flow-control valve and an expiratory valve or "exhalation manifold" (Fig. 3-5). In the simplest case, when inspiration is triggered on, the output control valve opens, the expiratory valve closes, and the only path left for gas is into the patient. When inspiration is cycled off, the output valve closes and the exhalation valve opens, flow from the ventilator ceases and the patient exhales out through the expiratory valve (see Fig. 3-2). The most sophisticated ventilators employ a complex interaction between the output flow-control valve and the exhalation valve, such that a wide variety of pressure, volume, and flow waveforms may be generated to synchronize the ventilator output with patient effort as much as possible.

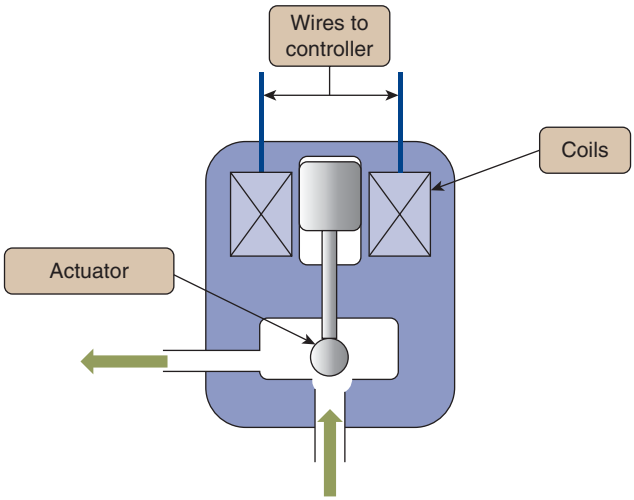


FIGURE 3-4 Schematic of an output flow-control valve.

FIGURE 3-6 Small, pneumatically powered transport ventilator using a pneumatic control system. (Reproduced, with permission, from CAREvent, O-Two Medical Technologies, Ontario, Canada.)

CONTROL SYSTEMS

In the simplest terms, the control system of a ventilator is comprised of components that generate the signals that operate the output valve and the exhalation manifold to obtain the desired output waveforms and modes of ventilation. Control systems may be based on mechanical, pneumatic, fluidic, or electronic components. Mechanical components include levers, pulleys, cams, and so on.[3] Pneumatic control circuits use gas pressure to operate diaphragms, jet entrainment devices, pistons, and other items. Use of lasers to create micro channels for gas flow has enabled miniaturization of ventilator control circuits that are powered entirely by gas pressure to create small, but sophisticated, ventilators for transport, such as the CAREvent (O-Two Medical Technologies) shown in Figure 3-6. Fluidic circuits are analogs of electronic logic circuits.[4] Just as an electronic logic circuit uses electricity, the fluidic circuit uses a very small gas flows to generate signals that operate switches and timing components. Both pneumatic and fluidic control systems are immune to failure from electromagnetic interference, such as around magnetic resonance imaging equipment. Examples of simple pneumatic and fluidic ventilator control circuits have been illustrated elsewhere.[5] By far, the majority of ventilators use electronic control circuits with microprocessors to manage the complex monitoring (e.g., from pressure and flow sensors) and control (valves) functions of modern ventilators used in almost every health care environment.

What makes one ventilator so different from another has as much to do with the control system software as it does with the hardware. The control software determines how the ventilator interacts with the patient; that is, the modes available. Thus, a discussion about control systems is essentially a discussion about mode capabilities and classifications. Chapter 2 describes the specific design principles of ventilator control systems in detail.

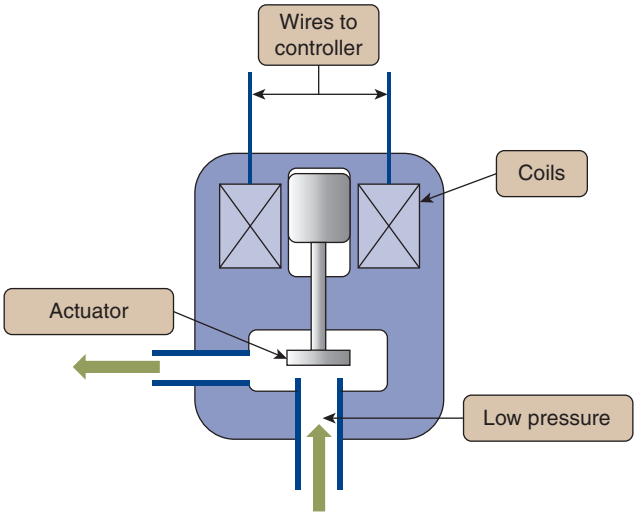


FIGURE 3-5 Schematic of an exhalation valve.

Outputs

Just as the study of cardiology involves the use of electrocardiograms and blood pressure waveforms, the study of mechanical ventilation requires an understanding of output waveforms. The waveforms of interest are the pressure, volume, and flow.

IDEALIZED PRESSURE, VOLUME, AND FLOW WAVEFORMS

Output waveforms are conveniently graphed in groups of three. The horizontal axis of all three graphs is the same and has the units of time. The vertical axes are in units of pressure, volume, and flow. For the purpose of identifying

characteristic waveform shapes, the specific baseline values are irrelevant. What is important is the relative magnitudes of each of the variables and how the value of one affects or is affected by the value of the others.

Figure 3-7 illustrates the typical waveforms available on modern ventilators. These waveforms are idealized; that is, they are precisely defined by mathematical equations and are meant to characterize the operation of the ventilator's control system. As such, they do not show the minor deviations, or "noise," often seen in waveforms recorded during actual ventilator use. This noise can be caused by a variety of extraneous factors such as vibration and flow turbulence. Of course, scaling of the horizontal and vertical axes can affect the appearance of actual waveforms considerably. Finally, the waveforms in Figure 3-7 do not show the effects of the resistance and compliance of the patient circuit.

No ventilator is an ideal pressure, volume, or flow controller, and ventilators are designed to only approximate a particular waveform. Idealized waveforms as shown in Figure 3-7 are, nevertheless, helpful because they are used commonly in other fields (e.g., electrical engineering), which makes it possible to use mathematical procedures and terminology that already have been established. For example, a standard mathematical equation is used to describe the most common ventilator waveforms for each control variable. This known equation may be substituted into the equation of motion, which is then solved to get the equations for the other two variables. Once the equations for pressure, volume, and flow are known, they are easily graphed. This is the procedure that was used to generate the graphs in Figure 3-7.

EFFECTS OF THE PATIENT CIRCUIT

The pressure, volume, and flow the patient actually receives are never precisely the same as what the clinician sets on the ventilator. Sometimes these differences are caused by instrument inaccuracies or calibration error. More commonly, the patient delivery circuit contributes to discrepancies between the desired and actual patient values. This is so because the patient circuit has its own compliance and resistance. Thus, the pressure measured inside a ventilator upstream of the patient always will be higher than the pressure at the airway opening because of patient circuit resistance. In addition, the volume and flow coming out of the ventilator's exhalation manifold will exceed those delivered to the patient because of the compliance of the patient circuit.

Exactly how the mechanical properties of the patient circuit affect ventilator performance depends on whether they are connected in series or in parallel with the patient. It turns out that the resistance of the patient circuit is connected in series whereas the compliance is modeled as a parallel connection. To understand this, we first make the simplifying assumption that we can examine the patient circuit's resistance separate from its compliance. It is intuitively obvious that the same flow of gas that comes from the ventilator travels through the circuit tubing as through the patient's airway opening. We also can see that the pressure drop across the patient circuit will be different from that across the respiratory system because they have different resistances. By a definition we borrow from electronics, when two circuit components share the same flow but have different pressure drops, they are connected in series. This

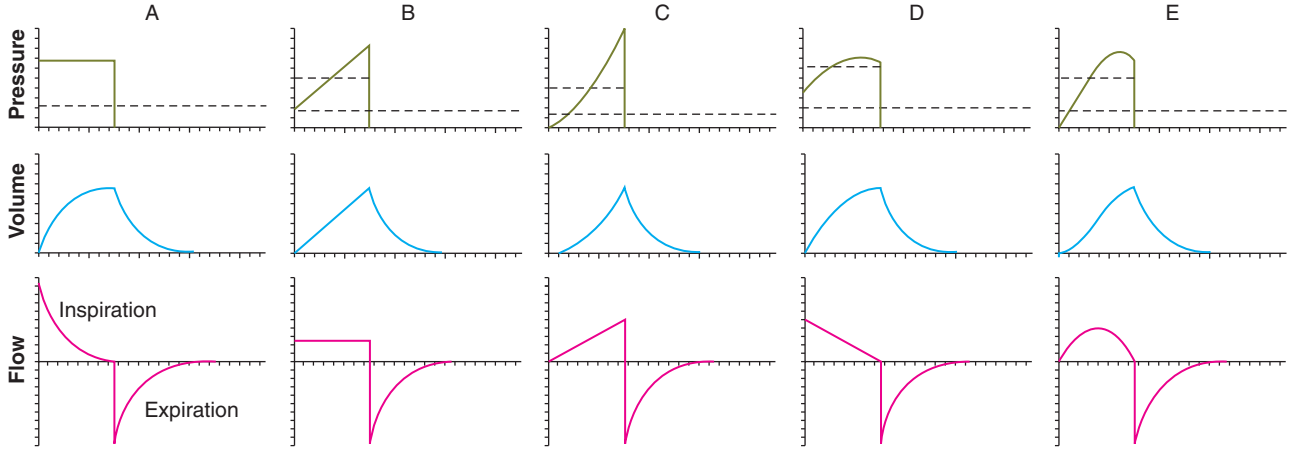


FIGURE 3-7 Idealized ventilator output waveforms. **A.** Pressure-controlled inspiration with a rectangular pressure waveform. **B.** Volume-controlled inspiration with a rectangular flow waveform. **C.** Volume-controlled inspiration with an ascending-ramp flow waveform. **D.** Volume-controlled inspiration with a descending-ramp flow waveform. **E.** Volume-controlled inspiration with a sinusoidal flow waveform. The *short dashed lines* represent mean inspiratory pressure, and the *long dashed lines* represent mean pressure for the complete respiratory cycle (i.e., mean airway pressure). Note that mean inspiratory pressure is the same as the pressure target in A. These waveforms were created as follows: (a) defining the control waveform using a mathematical equation (e.g., an ascending-ramp flow waveform is specified as flow = constant × time), (b) specifying the tidal volume for flow-control and volume-control waveforms, (c) specifying the resistance and compliance, (d) substituting the preceding information into the equation of motion for the respiratory system, and (e) using a computer to solve the equation for the unknown variables and plotting the results against time. (Reproduced, with permission, from Chatburn RL. *Fundamentals of Mechanical Ventilation.* Cleveland Heights, OH: Mandu Press; 2003:143.)

means that the patient circuit resistance, however small, adds to the total resistive load seen by the ventilator. Thus, in a volume-controlled breath, the peak inspiratory pressure is higher, and in a pressure-controlled breath, the tidal volume and peak flow are lower. In practice, the effect of patient circuit resistance is usually ignored because it is so much lower than the resistance of the respiratory system.

Now consider the patient circuit compliance. The effective compliance of the patient circuit is a combination of the tubing compliance and the compressibility of the gas inside it. As the ventilator delivers the breath to the patient, pressure at the airway opening rises relative to atmospheric pressure, which is the driving force for flow into the lungs. The patient circuit is connected between the ventilator and the airway, so the pressure it experiences across its walls is the same as that experienced by the respiratory system (remember that we are ignoring its resistance now, so we can ignore any pressure drop between the ventilator outlet and the airway opening). The volume change of the patient circuit tubing is different from that of the respiratory system because the compliance of the circuit is different. Because the patient circuit and the respiratory system fill with different volumes during the same inspiratory time, the flows they experience are different (remember that flow = volume ÷ time). Again borrowing a definition from electronics, if two circuit components share the same pressure drop but different flows, they are connected in parallel. Because they are in parallel, the two compliances are additive, so the total compliance is greater than either component.

Patient circuit compliance sometimes can be greater than respiratory system compliance and thus can have a large effect on ventilation. It must be accounted for either automatically by the ventilator or manually by increasing the tidal volume. For example, when ventilating neonates, patient circuit compliance can be as much as three times that of the respiratory system, even with small-bore tubing and a small-volume humidifier. Thus, when trying to deliver a preset tidal volume during volume-controlled ventilation, as little as 25% of the set volume will be delivered to the patient, with 75% compressed in the patient circuit. The compliance of the patient circuit can be determined by occluding the tubing at the patient *Y*, delivering a small volume under flow control (using zero positive end-expiratory pressure [PEEP]), and noting the resulting pressure. Using a short inspiratory hold will make it easier to read the pressure. Then compliance is calculated as before, by dividing the volume by the pressure. Once the patient circuit compliance is known, the set tidal volume can be corrected using the following equation:

$$V_{delivered} = \frac{V_{set}}{1+(C_{PC}/C_{RS})} \tag{1}$$

where $V_{delivered}$ is the tidal volume delivered to the patient, V_{set} is the tidal volume setting on the ventilator, C_{PC} is the patient circuit compliance, and C_{RS} is the respiratory system compliance.

We can get a more intuitive understanding of this equation if we put in some values. Suppose, for example, that we use the perfect patient circuit that has zero compliance. Substituting zero for C_{PC}, we get

$$\begin{aligned} V_{delivered} &= \frac{V_{set}}{1+(C_{PC}/C_{RS})} = \frac{V_{set}}{1+(0/C_{RS})} \\ &= \frac{V_{set}}{1+0} = \frac{V_{set}}{1} = V_{set} \end{aligned} \tag{2}$$

which shows that there is no effect on the delivered tidal volume. Suppose now that C_{PC} is as large as C_{RS} (i.e., $C_{PC} = C_{RS}$). Now we have

$$V_{delivered} = \frac{V_{set}}{1+(C_{PC}/C_{RS})} = \frac{V_{set}}{1+1} = \frac{V_{set}}{2} \tag{3}$$

in which case, half the volume from the ventilator goes to the patient, and the other half is compressed in the patient circuit. Some ventilators automatically compensate for gas lost to the patient circuit.[2]

The effect of the patient circuit is more troublesome during volume-controlled modes than during pressure-controlled modes. This is so because during volume control, the ventilator meters out a specific volume of gas, and unless it measures flow at the airway opening, it has no way of knowing how much goes to the patient and how much goes to the patient circuit. In contrast, during pressure-controlled modes, the ventilator simply meters out a set pressure change no matter where the gas goes. Because the respiratory system and the patient circuit compliance are in parallel, they both experience the same driving pressure (peak inspiratory pressure minus end-expiratory pressure), so tidal volume delivery is affected very little. The only effect might be that the patient circuit compliance may tend to increase the pressure rise time, which would tend to decrease peak flow and tidal volume slightly.

Another area where patient circuit compliance causes trouble is in the determination of auto-PEEP. There are several methods for determining auto-PEEP. One method to determine auto-PEEP during mechanical ventilation is to create an expiratory hold manually (i.e., delay the next inspiration) until static conditions prevail throughout the lungs (i.e., no flow anywhere in the lungs). The pressure at this time (total PEEP) minus the applied PEEP is an estimation of global auto-PEEP. Note that auto-PEEP may vary throughout the lungs depending on the distribution of lung disease and may not reflect pressure behind collapsed areas in patients with severe flow limitation. Auto-PEEP is an index of the gas trapped in the system at end expiration secondary to an insufficient expiratory time:

$$\text{measured auto-PEEP} = \frac{V_{trapped}}{C_{total}} \tag{4}$$

where $V_{trapped}$ is the volume of gas trapped in the patient and the patient circuit at end-expiration (above that associated with applied PEEP), and C_{total} is the total compliance of the

respiratory system and the patient circuit. The problem is that we want auto-PEEP to reflect the gas trapped in the patient, not in the circuit. If we know the compliances of the patient circuit and the respiratory system, we can correct the measured auto-PEEP as follows:

$$\text{true auto-PEEP} = \frac{C_{RS} + C_{PC}}{C_{RS}} \times \text{measured auto-PEEP} \quad (5)$$

where true auto-PEEP is that which exists in the lungs, measured auto-PEEP is the amount of end-expiratory pressure in equilibration with the lungs and the patient circuit, C_{RS} is the respiratory system compliance, and C_{PC} is the patient circuit compliance. If the ventilator displays auto-PEEP on its monitor, check the ventilator's operating manual to see whether or not the auto-PEEP calculation is corrected for patient circuit compliance. The larger C_{PC} is relative to C_{RS}, the larger will be the error. Again, the error will be most noticeable in pediatric and neonatal patients.

THE OPERATOR INTERFACE

The operator interaction with the ventilator mainly happens through the ventilator display. The display or interface has evolved in parallel with the ventilators. The key to this evolution are the technological advances in the last three decades.[2] The microprocessors, the digital displays, and the interactive screens have all permeated from other technological advances into the ventilator world. There are still remnants of the evolutionary process. In their initial ventilator generations, the interface had no or minimal manifestation of the interaction with the patient. The operator would enter the ventilator settings by using knobs or buttons that regulated simple functions (pressure, flow, or time). The results of these changes were evaluated in the patient clinical response, and occasionally through simple pressure analog displays. Some ventilators still use these type of displays (e.g., CareFusion 3100A high-frequency oscillator and Puritan Bennett LP-10, Fig. 3-8).

Most of the ventilators produced in the last decade have advanced displays, including liquid crystal displays and color touch screens with one or more multipurpose knobs or buttons. This allows the user to scroll through different menus and to select and activate the selections (e.g., Hamilton G5 ventilator, Fig. 3-9). The operator can customize the screen to the operator's needs. Current ventilators allow graphical displays of alarms, settings, respiratory system calculations, and measurements. The ventilator display evolution has not necessarily resulted in easier management of the ventilator. These advances brought issues with the amount of information displayed, the actions taken with that information, and the ease of use of certain interfaces.[6] As the level of sophistication has increased, we have been able to increase the number of ventilation parameters monitored. This requires a new level of training and understanding of human behavior. For example, a mode of ventilation may be preferentially chosen based on the amount of alarms it triggers,[7] or its ease of use.[6,8]

FIGURE 3-8 Puritan Bennett LP-10 home-care ventilator. (Image with permission from Nellcor Puritan Bennett LLC, Boulder, Colorado, doing business with Covidien.)

Operator Inputs

The operator input refers to parameters or settings entered by the operator of the ventilator. Each mode of ventilation has particular features, some of which can be adjusted by the operator. We describe here the most common adjustable parameters. The effect of each parameter on the lung is better understood under the light of the equation of motion (see Chapter 2).[9,10] A change of one parameter will lead to changes in others (i.e., in volume control, for the same respiratory characteristics changing the tidal volume will cause a change in peak airway pressure). Furthermore, knowing the basic construction and characteristics of a

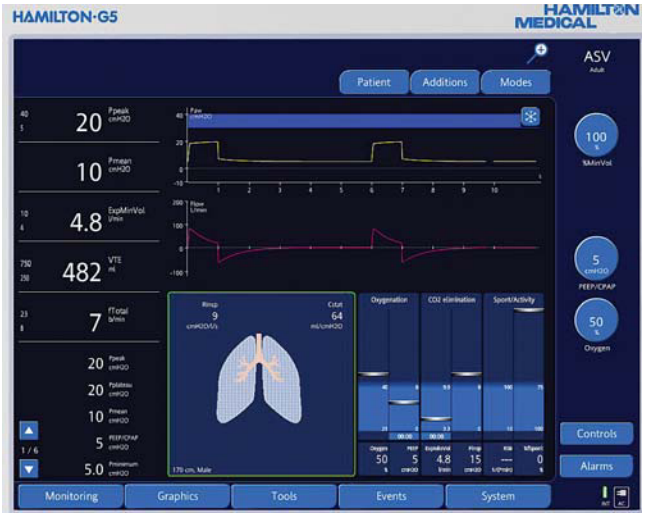


FIGURE 3-9 G5 ventilator. (Reproduced with permission from Hamilton Medical, AG.)

mode of ventilation (volume vs. pressure control breaths) or the breath sequence (mandatory vs. spontaneous) will help understand how the setting will affect the ventilator output (see Chapter 2).

The operator input is presented below in the order that follows the progression of a breath; starting with the gas inhaled, to triggering, targeting, cycling, and baseline variables.

Inspired Gas Concentration

A mechanical ventilator has the capacity of delivering different mixtures of gas. Most ventilators allow the administration of specific concentrations of oxygen. A few allow the administration of helium, nitric oxide, or anesthesia gases.

OXYGEN

Oxygen is the most common gas administered to patients undergoing mechanical ventilation. The oxygen percentage in the inspired gas (FI_{O_2}) can be regulated in most ventilators by means of a direct adjustment of a specific control (21% to 100%). However, this is not true for all ventilators. For example, some home ventilators (e.g., LP-10 or the LTV 1150, Pulmonetic, CareFusion) use a connection to a low-pressure oxygen source to the ventilator or the patient circuit. The following formula can calculate the flow of oxygen to achieve a desired oxygen concentration:

$$O_2 \text{ required} = \frac{f \times V_T \times (\text{desired } FIo_2 - 0.21)}{0.79} \qquad (6)$$

where O_2 required is 100% oxygen flow in L/min, f is the breathing frequency in breaths/min, V_T is the tidal volume in liters and the FI_{O_2} is the patient O_2 concentration desired in decimal format (i.e., 30% = 0.3). An oxygen analyzer should be used to confirm the measurements. It must be recognized that changes in oxygen flow, breathing rate, or tidal volume will change the FI_{O_2}.

When transporting the critically ill patient, availability of oxygen supplies for the mechanically ventilated patient is crucial. Size and weight of cylinders makes transport difficult and presents an increased risk of fire. Branson et al. have described a solution using a portable oxygen concentrator (SeQual Eclipse II) paired with the Impact 754 and Pulmonetics LTV-1200 ventilators.[11]

For the rest of the current mechanical ventilators, the ventilator adjusts the mixture of air and oxygen to achieve the desired FIo_2. The mixing of air is achieved by an internal or external blender. A blender may use proportioning valves that regulate the flow of air and oxygen to a mixing changer (Fig. 3-10). It is similar to the mechanism used to mix hot and cold water in a shower—the more oxygen needed, the larger the opening for oxygen and the smaller it is for air. To work properly, the blender requires a constant pressure within the working ranges of the device.

Most current ventilators have oxygen sensors to monitor the FI_{O_2}. The oxygen sensor gives feedback to the operator to adjust the mixture, or alarms if there is a discrepancy between the set and delivered FI_{O_2}. The oxygen sensors detect changes in electrical current, which is proportional to the oxygen concentration. The most common techniques are: (a) paramagnetic, (b) polarographic, and (c) galvanic.[12]

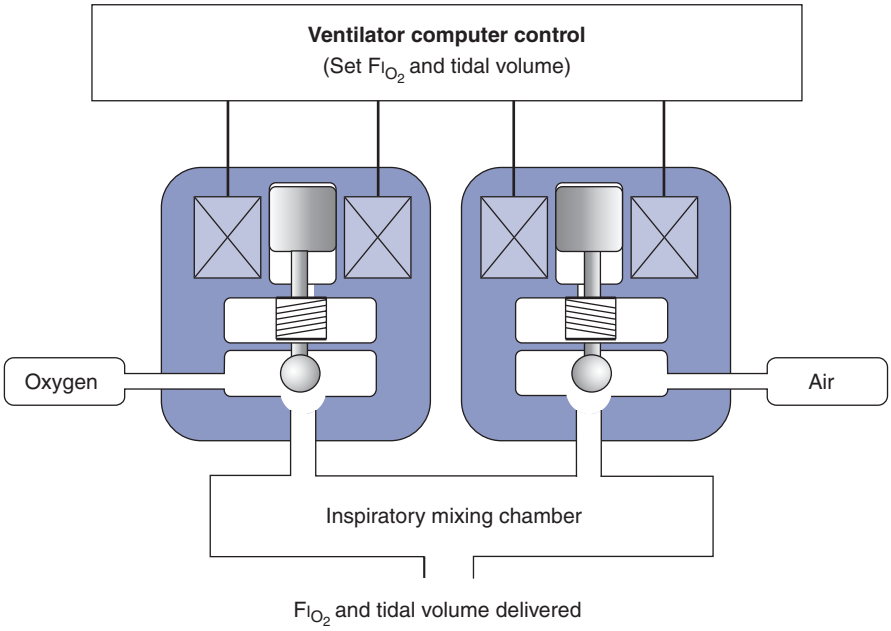


FIGURE 3-10 Schematic of a ventilator air–oxygen blending system using proportional valves.

TABLE 3-1: PROPERTIES OF PURE GASES AND AIR

Gas	Thermal Conductivity (κ) (μcal · cm · s · °k)	viscosity (η) (Micropoises)	Density (ρ) (g/L)
Helium (He)	352.0	188.7	0.1785
Nitrogen (N_2)	58.0	167.4	1.251
Oxygen (O_2)	58.5	192.6	1.429
Air	58.0	170.8	1.293

HELIOX

Mixtures of helium and oxygen (heliox, HeO_2) instead of air and oxygen are occasionally used to help patients on mechanical ventilation with obstructive airway diseases. Helium is less dense than air (Table 3-1).[13] The decrease in density interferes with flow measurements, inspiratory and expiratory valve accuracy, and gas mixing.[14] Several studies have evaluated the performance of mechanical ventilators delivering heliox[14–16] and have shown that heliox does affect the performance of the ventilator. The interference of heliox is more evident in volume-control modes than in pressure-control modes.[14,17] In pressure-control mode, the ventilator targets a set inspiratory pressure and the delivered tidal volume is dependent only on the mechanical properties of the respiratory system. The time constant may decrease but the delivered volume should be the same as for nonheliox gas delivery. In volume-control mode, delivered volume may be larger than, smaller than, or the same as expected depending on the design of the ventilator.[14] Only a few ventilators (Maquet Servo *i* with heliox option, Hamilton G5 with heliox option, and the Viasys Avea with comprehensive model) are designed and calibrated for heliox delivery. Otherwise, the operator needs to be aware of the specific ventilator performance and correction formulas and factors[14] such that potentially hazardous conditions do not develop.

NITRIC OXIDE

Inhaled nitric oxide (NO) is used as selective pulmonary vasodilator for patients with pulmonary hypertension, life-threatening hypoxia, or right-heart failure. Different devices to deliver NO have been described in the literature. Most of them were custom made and required the use of mixing chambers, stand-alone NO/nitric dioxide monitors, and manual titration of the gas flow. The large amount of custom-made devices led to inconsistent administration of NO.[18] In 1998, the American Society for Testing Materials (ASTM) committee on anesthetic and respiratory equipment developed a standard to provide a minimum degree of safety of the devices used to deliver NO. The recommendation was to use a NO administration apparatus, and a NO/nitrogen dioxide analyzer. The Food and Drug Administration (FDA) enforces this recommendation, and so far, only one device is approved in the United States. The INOvent (Ikaria Inc, Clinton, NJ) delivery system uses a closed-loop scheme to measure and deliver NO in proportion to the inspiratory flow from the ventilator. NO is injected in the proximal limb of the inspiratory circuit, and measured close to the connection between the patient circuit and the endotracheal tube. Two portable systems are available—INO Max DS (Ikaria) and AeroNOx (PulmoNOx, Alberta, CA). As these devices are not universally available, the following formula[19] can be used to calculate the NO flow rate required to achieve a desired concentration of NO when injected in the inspiratory limb at a constant gas flow,

$$Q_{NO} = \left(\frac{C_{NOset}}{C_{NOcyl} - C_{NOset}}\right) \times Q_V \qquad (7)$$

where Q_{NO} is the flow rate of nitric oxide in L/min, C_{NOset} is the desired NO concentration in parts per million (ppm), C_{NOcyl} is the NO concentration in the cylinder in ppm (usually 800 ppm) and the Q_V is the ventilator gas flow.

The formula is accurate for constant flow systems. This presents a major problem when used with intermittent breaths (as most modes of ventilation) the patient will receive variable amounts of NO (a "bolus" with each mechanical breath).[20] Furthermore, whenever the ventilator settings or the patient breathing pattern changes, the NO delivery will change. Finally, the use of NO will alter the gas delivery of the ventilator. For example, the INOvent system will add gas to and extract gas from the delivered breath. At 80 ppm it adds 10% more gas, although it also withdraws 230 mL/min through the gas-sampling port. Thus, the oxygen delivered will decrease, and the tidal volume may increase. The changes seem to be small (unless you see it in pediatric proportions), but it may affect the ventilator's performance. Furthermore, as a flow of gas is introduced, the flow-triggering performance may be affected.

Trigger Variables

A ventilator-assisted breath can be started (triggered) by the machine or the patient. A machine-triggered breath is defined by the start of the inspiratory phase *independent*

of any signal from the patient. The operator typically sets a breath frequency for machine-triggered breaths. A patient-triggered breath is one for which inspiration is started solely by a signal from the patient. The key operator set variable for patient triggering is sensitivity, or the magnitude of the patient signal required to initiate inspiratory flow. The patient signal can be obtained from measuring the airway pressure, flow, volume, electromyogram (EMG),[21] abdominal motion (Graseby capsule[22]), thoracic impedance,[23] or any other measurable signal of respiratory activity.[24] Most intensive care ventilators measure pressure and flow (volume is integrated from flow) at the circuit. There are only a few ventilators that use other sources of signaling, diaphragmatic EMG (Servo *i* NAVA), thoracic impedance (Sechrist SAVI), and abdominal motion (Infant Star STAR SYNC, which is no longer commercially available).[24,25]

Ventilator triggering characteristics can be evaluated using different metrics.[23,26–28] The most sophisticated device for evaluating ventilator performance is the ASL lung simulator (IngMar Medical Ltd., Pittsburgh, PA). This device can simulate both passive lung mechanics (e.g., resistance and compliance) as well as patient inspiratory and expiratory effort. It can display and record pressure, volume, and flow signals, and calculate a wide variety of performance metrics. Figure 3-11 shows an example of these waveforms with specific reference points for calculating performance metrics (from operator's manual for software version 3.2). Using these reference points we can define the following key trigger metrics: P_{min} (maximum pressure drop relative to PEEP during the trigger phase), pressure-time product (*Paw–PEEP dt* from start of effort to return of airway pressure [Paw] to PEEP), patient trigger work (*Paw–PEEP dv* from start of effort to return of Paw, to PEEP), and time to trigger (period from the start of effort to the return of Paw to PEEP).

TIME

Time is measured by the internal ventilator processor. The next breath is time triggered (in the absence of a patient trigger event) when the expiratory time has reached the threshold to maintain a set respiratory rate (e.g., if the set rate is 10 breaths per minute and the inspiratory time is set at 1 second, then the expiratory time is 5 seconds). Some modes allow the user to set the inspiratory and expiratory time [e.g., airway pressure release ventilation (APRV) and biphasic], thus fixing the inspiratory-to-expiratory timing (*I:E*) ratio and respiratory rate. In an effort to improve patient–ventilator interactions, the ventilator may synchronize the mandatory breath with the patient's triggering signal if it falls within a threshold. The classic example is synchronized intermittent mandatory ventilation (SIMV). More recently APRV, as programmed in the Evita XL, delivers a machine breath if the patient trigger signal falls within 25% of the triggering time.[29] Time triggering is also found as a safety mechanism. The operator or manufacturer enters a time after which the apnea alarm will trigger the delivery of a preset breath after a preset time is reached.

PRESSURE

The patient inspiratory effort causes a drop in pressure in the airway and the circuit. Inspiration starts when pressure falls below the preset "sensitivity" threshold. The site of measurement will have an impact on the performance of the device. Pressure signals travel at the speed of sound, approximately 1 ft/ms.[30] The farther the sensor is from the signal source, the longer the potential time delay. The closest measurements can be done in the trachea. Tracheal pressure measurements reflect actual airway pressure as the endotracheal tube resistance is bypassed. When used for ventilator

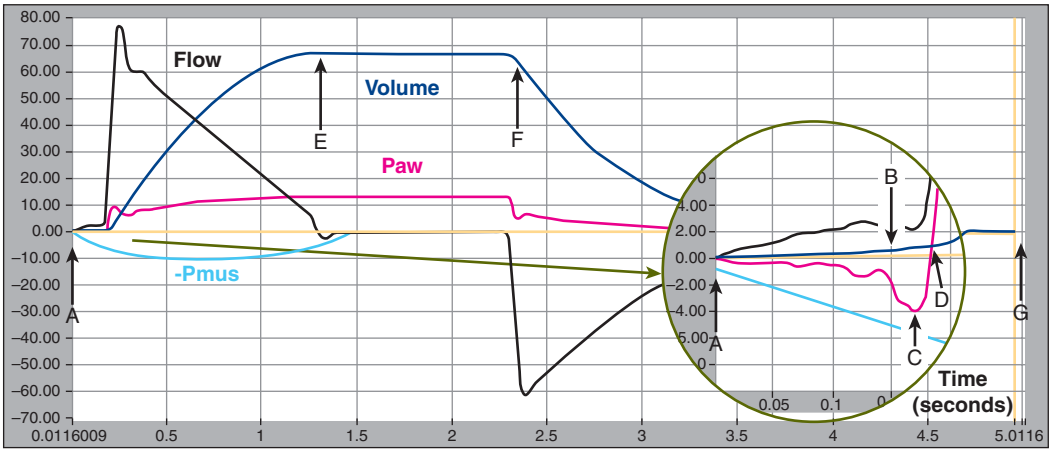


FIGURE 3-11 Reference points on pressure, volume, and flow waveforms recorded by the ASL 5000 (IngMar Medical Ltd, Pittsburgh, PA). **A.** Start of inspiratory effort, **B.** beginning of inhalation as determined by the "breath start volume threshold," **C.** lowest pressure during the trigger phase, Pmin, **D.** return of airway pressure to baseline during the trigger phase, **E.** end of inspiratory time, i.e., negative-going zero flow crossing, **F.** beginning of exhalation as determined by the "expiratory start volume threshold," and **G.** end of expiratory time, i.e., positive-going zero flow crossing. (Reproduced, with permission, from Ingmar Medical. *ASL 5000 v3.2 Operator's Manual.* Pittsburgh, PA: Author.)

TABLE 3-2: ADVANTAGES AND DISADVANTAGES OF THE DIFFERENT CIRCUIT PRESSURE-SENSING SITES

Advantages	Disadvantages
A. Exhalation port: Well protected from mechanical abuse. During mechanical inhalation, accurately reads pressure at the Y. During inhalation, increases in inspiratory or expiratory circuit resistance do not compromise inspiratory flow output, except for manyfold increases.	Requires protection from moisture of exhaled gas. During spontaneous inspiration, underestimates pressure generated at the Y to trigger the ventilator. During exhalation, underestimates pressure at the Y. During exhalation, increases in expiratory circuit resistance compromise expiratory flow. Hence, system requires well-maintained expiratory filter to ensure that expiratory circuit resistance remains low.
B. Inhalation port: Well protected from mechanical abuse. Does not require protection from moisture or additional filters. During exhalation, accurately reads pressure at the Y as long as the inspiratory circuit remains patient. During inhalation, increases in expiratory circuit resistance do not compromise inspiratory-flow output.	During mechanical inhalation, overestimates pressure at the Y. During spontaneous inspiration, underestimates pressure generated at the Y to trigger the ventilator. During inhalation, increases in inspiratory circuit resistance compromise inspiratory flow output. For example, factors such as selection of humidifier and type of patient circuit yield varying patient inspiratory efforts for fixed ventilator settings.
C. Patient Y: During inhalation and exhalation, accurately reads both inspiratory and expiratory pressures. Pressure readings reflect relative condition of inspiratory and expiratory circuits.	Susceptible to mechanical abuse. Requires a separate pressure-sensing tube, which is prone to occlusion, blockage, and disconnection, all of which prevent sensing of patient effort.

Source: Modified, with permission, from Sassoon CSH. Mechanical ventilator design and function: the trigger variable. *Respir Care*. 1992;37:1056–1069.

triggering, tracheal pressure sensing results in decreased work of breathing.[31–33] However, tracheal pressure measurements are not routinely done and require special equipment (endotracheal tube with monitoring port) and no current ventilator uses it to routinely trigger the ventilator.

The other sites of pressure measurement are the patient circuit Y or at the inspiratory or expiratory ports, each with its advantages and disadvantages (Table 3-2). Trigger performance will also be affected by the presence of humidifiers, filters, water condensation, patient circuit and exhalation valves. These will most often dampen, or rarely amplify, the pressure signal. Clinically, the presence of a dampened signal will require a larger pressure change (higher work of breathing) to reach the trigger threshold. On the contrary, presence of water in the pressure tubing may cause oscillation, which can falsely trigger mechanical breaths.

The trigger pressure sensitivity is usually set at 0.5 to 1.5 cm H_2O below the baseline pressure. Common practice is to increase the sensitivity (i.e., decrease the pressure drop) until autotriggering occurs and then reduce sensitivity until the autotriggering just stops.[30] Note that each ventilator comes with predetermined manufacturer set values and can be adjusted.

FLOW

Flow triggering is based on the detection of a change in a constant, small, baseline (bias) flow through the patient circuit. The operator sets a flow sensitivity threshold. When the change in flow reaches the threshold, a breath is delivered. The changes in flow are detected at the expiratory valves or by a flow sensor in the patient circuit. The ventilator measures the flow from the ventilator and from the patient. In a closed circuit, the two flow values should remain equal in the absence of patient effort. When the patient makes an inspiratory effort, the expiratory flow drops, creating a difference between the inspiratory and expiratory flow values. When the difference in values reaches the preset sensitivity threshold, a breath is delivered. Some systems (Puritan Bennett, 7200) allow the operator to set both the bias flow and the trigger sensitivity. Newer devices set the bias flow according to the operator selected value for the triggering sensitivity. For example, the Puritan Bennett 840 sets the flow 1.5 L/min above the selected sensitivity, and the Hamilton G5 automatically sets the bias flow equal to two times the set sensitivity threshold. As a backup, if flow sensor is kinked or taken out of line, an internal pressure trigger of –2 cm H_2O is used until the flow sensor is "online" again.

Flow change may be detected by placing a sensor just before the endotracheal tube. The close proximity to the patient may enhance triggering. It, however, exposes the sensor to secretions and moisture, which may affect its performance. Flow triggering seems more efficient than pressure triggering in terms of work of breathing.[34] This, however, seems of no particular clinical relevance in the presence of appropriately set pressure triggering.[35] Flow sensing may cause autotriggering secondary to noninspiratory flow changes. The flow change can happen in either the ventilator circuit (leak in the circuit or endotracheal tube) or the patient (cardiogenic oscillations or bronchopleural fistula).[36,37]

A novel approach to flow triggering is offered on the Dräger Infinity V500 ventilator in the APRV mode. Rather than setting a T-low time to determine the time triggering of each mandatory breath, the operator may set a percent of peak *expiratory* flow as the trigger threshold.

VOLUME

A breath may be triggered when a preset volume is detected as the result of a patient inspiratory effort. This is similar to

flow triggering but using volume has the theoretical advantage of being less susceptible to signal noise (i.e., integrating flow to get volume cancels out some noise because of flow oscillations). Volume triggering is rare in ventilators but can be found on the Dräger Babylog VN500 infant ventilator.

DIAPHRAGMATIC SIGNAL

The ideal approach to coordinate a mechanical ventilator with the patient inspiratory effort would be to use the neural output of the respiratory center. Direct measurement of the respiratory center output is currently not possible. The phrenic nerve has been used as a trigger signal in animal models,[38,39] but not in humans. The only available clinical approach is measurement of the diaphragmatic electrical activity (Edi). Because the Edi is an electric signal, it easily becomes contaminated by the electrical activity of the heart, the esophagus, and other muscles.[21] More importantly, the Edi requires an intact respiratory center, phrenic nerve, neuromuscular junction, and assumes that the diaphragm is the primary inspiratory muscle (e.g., rather than accessory muscles of ventilation).

The only clinically available system that uses diaphragmatic signal trigging is the neurally adjusted ventilatory assistance (NAVA) system. An esophageal catheter is used to measure the Edi. The sensitivity is set by entering a value above the background electrical noise. The trigger value is set in microvolts and represents the change in the electrical signal rather than an absolute value.[40] The default setting is 0.5 microvolts, but it can be adjusted from 0 to 2 microvolts. As a backup trigger signal in the absence of a measurable Edi, NAVA uses flow or pressure triggering, whichever happens first.

OTHER SIGNALS

The BiPAP Vision (Respironics Inc., Murrysville, PA) uses a triggering mechanism called *shape-signal*. The ventilator microprocessor generates a new flow signal, which is offset from the actual flow by 0.25 L/s and delays it for 300 milli seconds. The delay causes the flow shape signal to be slightly behind the patient's flow rate. The mechanical breath is triggered when a sudden decrease in expiratory flow from an inspiratory effort crosses the shape signal.[41]

The Sechrist SAVI system (Sechrist Industries, Anaheim, CA) is the only mode available that uses transthoracic electrical impedance to trigger the ventilator.[25] The thoracic impedance is obtained by placing two chest leads, one in the anterior axillary line on the right and the other in the posterior axillary line on the left. The sensors are placed high enough to avoid costal and subcostal retractions. The chest sensors measure the electrical impedance across the human body. As a breath occurs, the transthoracic impedance changes as a result of a different ratio of air-to-fluid in the thorax. The triggering threshold can be adjusted. The cardiac cycle may also cause interference with the signal.[22,25]

Target Variables

During inspiration, the variable limiting the magnitude of any parameter is called the target variable (previously known as the limit of the control variable, but the term *limit* is now reserved for alarm and safety conditions rather than control settings).[42]A target is a predetermined goal of ventilator output. Targets can be viewed as the parameters of the targeting scheme (see Chapter 2). *Within-breath targets* are the parameters of the pressure, volume, or flow waveform. Examples of within-breath targets include inspiratory flow or pressure rise time (set-point targeting), inspiratory pressure and tidal volume (dual targeting), and constant of proportionality between inspiratory pressure and patient effort (servo targeting). *Between-breath targets* serve to modify the within-breath targets and/or the overall ventilatory pattern. Between-breath targets are used with more advanced targeting schemes, where targets act over multiple breaths. A simple example of a between-breath target is to compare actual exhaled volume to a preset between-breath tidal volume so as to automatically adjust the within-breath constant pressure or flow target for the next breath. Examples of between-breath targets and targeting schemes include average tidal volume (for adaptive targeting), percent minute ventilation (for optimal targeting), and combined partial pressure of carbon dioxide, volume, and frequency values describing a "zone of comfort" (for intelligent targeting).

PRESSURE

The ventilator uses microprocessors to control the delivery of pressure. The pressure can be delivered with any pressure profile and in response to many signals. Currently, most modes of ventilation in which inspiratory pressure is targeted deliver the pressure rapidly and attempt to maintain the pressure constant throughout the inspiratory phase (square waveform). This means that the performance of the ventilator depends on the delivery of the pressure waveform and any departure from the ideal waveform leads to differences in performance between ventilators.[43,44]

Inspiratory Pressure. The pressure rise during inspiration associated with volume and flow delivery is set by the operator (pressure control–continuous mandatory ventilation) or closed-loop algorithms (e.g., pressure-regulated volume control). Care should be exercised while setting the ventilator or reading the literature as there is significant variability between ventilator manufacturers and peer-reviewed literature in the definitions and nomenclature related to inspiratory pressures.[43] The main problem stems from what historically has been used to define the inspiratory pressure. For example, in the same ventilator, for

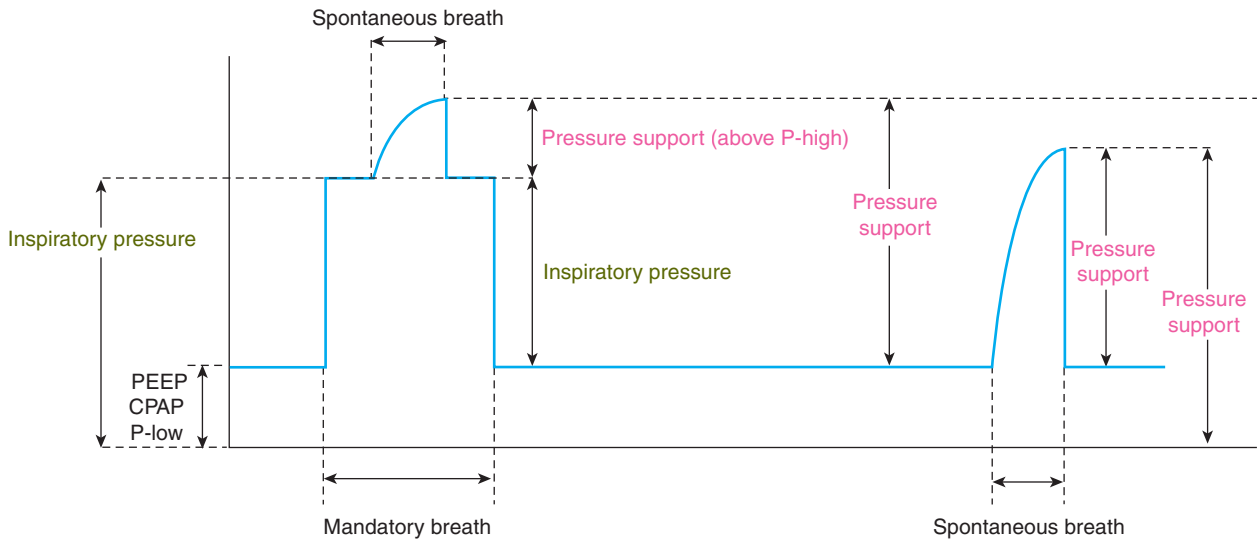


FIGURE 3-12 Idealized airway pressure waveform showing various conventions used for pressure parameters. Note that there are two ways to define inspiratory pressure for mandatory breaths (*green*) and four ways to define inspiratory pressure (i.e., pressure support) for spontaneous breaths (*red*). *CPAP*, continuous positive airway pressure; *PEEP*, positive end-expiratory pressure; *P-high*, high pressure; *P-low*, low pressure. (Reproduced, with permission, from Chatburn RL, Volsko TA. Documentation issues for mechanical ventilation in pressure-control modes. *Respir Care*. 2010;55(12):1705–1716.)

pressure control–continuous mandatory ventilation breaths the peak inspiratory pressure is stated in reference to the set end-expiratory pressure (PEEP), but for APRV the peak inspiratory pressure is stated in reference to the atmospheric pressure. To compound the confusion, on some ventilators the value of pressure support is set relative to PEEP (e.g., Drager Evita XL, Puritan Bennett 840), on others (LTV 950) pressure support is set relative to the atmospheric pressure (i.e., atmospheric pressure = zero airway pressure), and on at least one ventilator (BiVent in Servo *i*) pressure support may be set relative to inspiratory pressure (P-high). Figure 3-12 illustrates the two different ways used to define inspiratory pressure and the four different ways to define pressure support. Figure 3-13 illustrates the proposed solution to this problem.[43] In this proposal, the term *inspiratory pressure* is defined as the set change in airway pressure during inspiration relative to set end-expiratory airway pressure during pressure-control modes.

On some ventilators, inspiratory pressure rise is set relative to atmospheric pressure rather than set end-expiratory pressure. To distinguish this from inspiratory pressure as defined relative to PEEP, the term *peak inspiratory pressure* has been proposed.[43] In contrast "peak *airway* pressure" is the *measured* peak airway pressure relative to atmospheric pressure. Often, for a good pressure-control system, there is seemingly no difference between set peak inspiratory pressure and measured peak airway pressure on the airway-pressure waveform during pressure-control modes. And even if the operator sees a transient small difference, this is not considered clinically important in most nonalarm cases. This leads clinicians to conceptually oversimplify what they see and make the mistake of assuming inspiratory pressure and peak airway pressure are synonymous. For example, measured peak airway pressure is often higher than set peak inspiratory pressure because of pressure transients from an underdamped pressure-control system or noise from patient movement. The introduction of the so-called active exhalation valve made possible unrestricted spontaneous breaths during the inspiratory phase of a mandatory pressure-control breath. New modes brought new terms. For example, P-high or PEEP high refers to the peak inspiratory pressure above atmospheric pressure in APRV (again, there is no standardization of either terminology or symbology in this mode).

P_{max}. The Drager Evita XL, when set in volume-control modes, allows the operator to set the maximum pressure (P_{max}) that can be achieved during the delivery of a mandatory breath. The goal is to prevent pressure peaks while maintaining the set tidal volume. When the P_{max} is reached during a given inspiration, the ventilator switches from volume control to pressure control (dual targeting) using the P_{max} setting as the inspiratory pressure target. If the set tidal volume cannot be reached in the set inspiratory time, the ventilator will alarm.[45]

Rise Time. The speed with which the airway pressure reaches the set inspiratory pressure is called the *rise time*. (Rise time for flow can be set in the Maquet Servo *i*, but this feature is rare on ventilators.) The rise time may be set by the operator or automatically adjusted based on a computer algorithm (e500, Newport Medical Instruments Inc, Newport Beach, California). The name used to indicate pressure rise time varies by ventilator brand (e.g., inspiratory

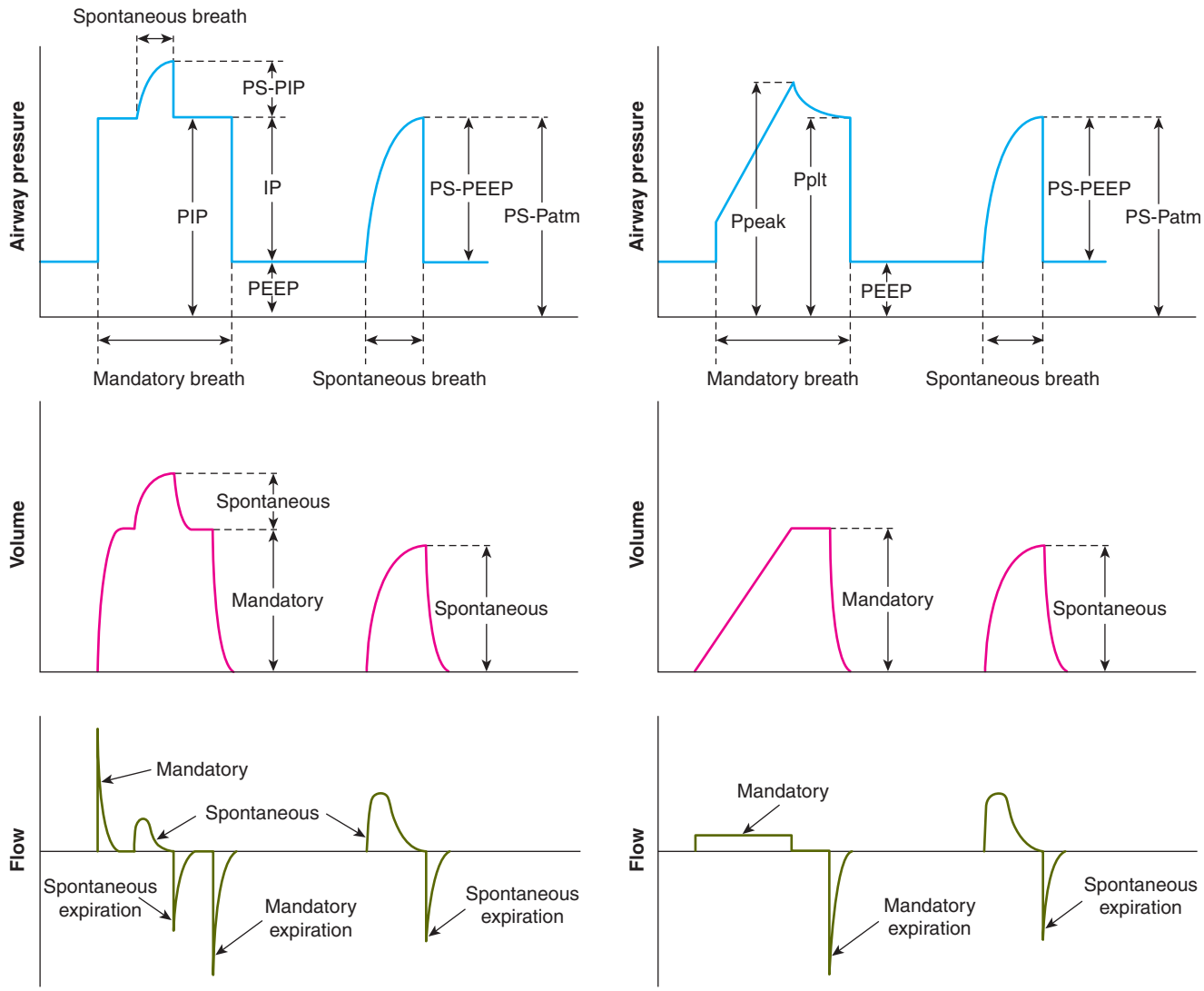


FIGURE 3-13 Idealized pressure, volume, and flow waveforms for pressure control and volume control illustrating the use of proposed conventions for both set and measured airway pressures. *IP*, inspiratory pressure; *PEEP*, positive end-expiratory pressure; *PIP*, peak inspiratory pressure; *Ppeak*, peak pressure; *Pplt*, plateau pressure; *PS-Patm*, pressure support relative to atmospheric pressure; *PS-PEEP*, pressure support relative to positive end expiratory pressure; *PS-PIP*, pressure support relative to peak inspiratory pressure. (Reproduced, with permission, from Chatburn RL, Volsko TA. Documentation issues for mechanical ventilation in pressure-control modes. *Respir Care*. 2010;55(12):1705–1716.)

slope, P-ramp, plateau%, and slope rise time). Adjusting the rise time influences the synchronization between the patient and the ventilator secondary to changes in the initial inspiratory flow rate. The lower the rise time, the faster the pressurization rate[46] and the higher the peak inspiratory flow.[47] A higher initial inspiratory flow rate may decrease the work of breathing but can lead to patient discomfort and worse patient–ventilator synchrony. Conversely, too slow a rise time may result in increased work of breathing and longer mechanical inspiratory time, leading to a dissociation between patient breathing effort and the mechanical breath. That is, the relation between work of breathing, respiratory drive, and comfort with the duration of the rise time is not proportional.[46,48] Because rules for setting an optimal rise time are lacking, based on these studies, both very rapid and slow rise time should be avoided. A more gradual rise may be needed in awake patients (for comfort) or patients with low compliance to prevent pressure overshoot and premature cycling of inspiration (Fig. 3-14).

TIDAL VOLUME

The operator is required to enter a tidal volume in any volume-control mode. This may be a direct setting or an indirect one by setting frequency or minute ventilation. The ventilator will control the tidal volume and the pressure will be the dependent variable. A tidal volume target, however, may also be set when the mode uses adaptive targeting in pressure control (e.g., pressure-regulated volume control [PRVC] on the Maquet ventilators).[49] In such a case, inspiratory pressure

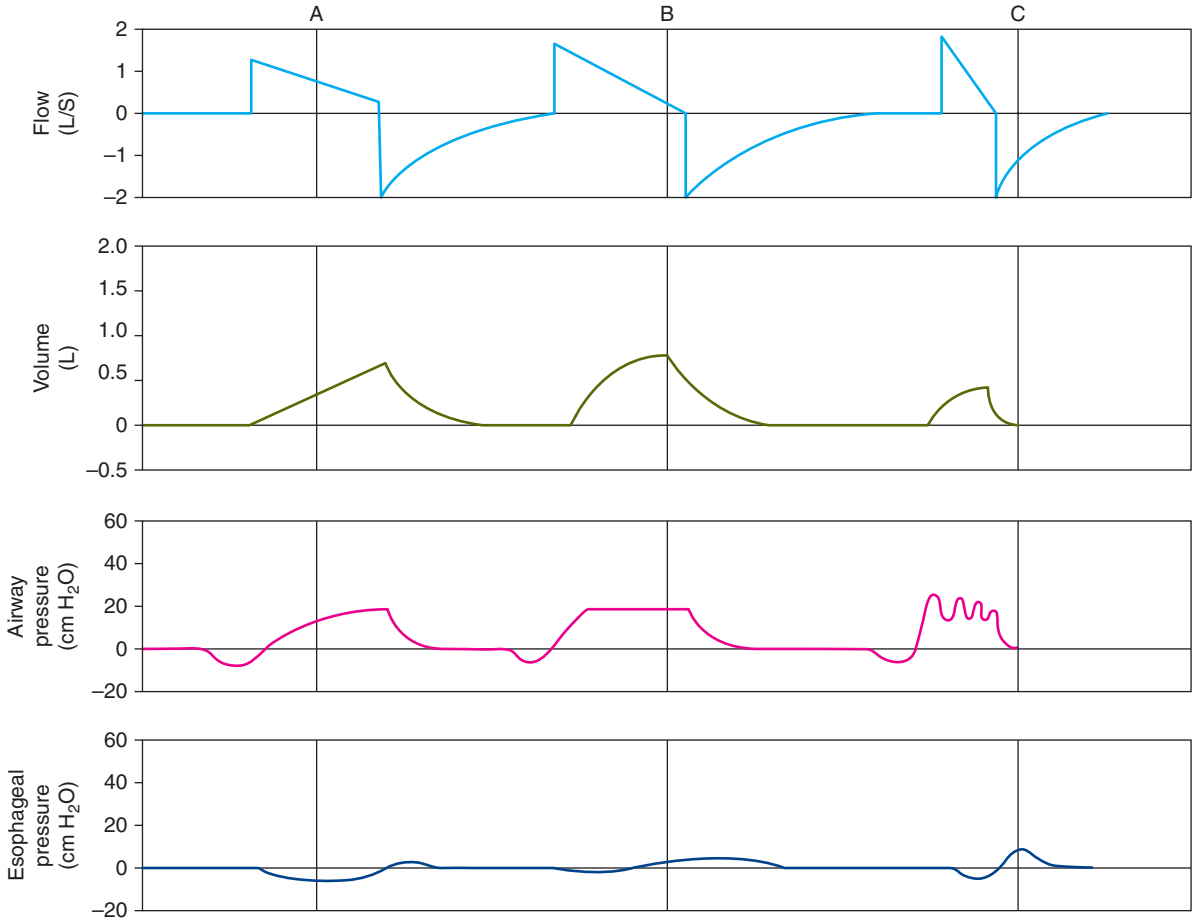


FIGURE 3-14 Examples of different pressure rise times in three breaths in pressure-support mode. **A.** Rise time is set very low, resulting in a lower peak inspiratory flow. **B.** Rise time is set higher, resulting in a higher peak flow and shorter inspiratory time. **C.** Rise time is set very high, resulting in "ringing" of airway pressure signal and peak flow that is uncomfortable to the patient, who exerts an expiratory effort and prematurely terminates inspiration (indicated by the positive deflection of esophageal pressure). (Reproduced, with permission, from Macintyre NR. Patient-ventilator interactions: optimizing conventional modes. *Respir Care*. 2011;56(1):73–81.)

is automatically adjusted between breaths by the ventilator to achieve an average measured tidal volume equal to the operator set target. There are four basic ways ventilators deliver a preset tidal volume (from least used to most commonly used):

1. By measuring the volume delivered and using the signal in a feedback control loop to manipulate the volume waveform.
2. By the displacement of a piston or bellows. An example of this is the Puritan Bennett LP10 home-care ventilator (piston) or some anesthesia ventilators (bellows).
3. By controlling the inspiratory pressure within a breath and automatically adjusting it between breaths to deliver a minimum set tidal volume. The volume delivered is targeted by a closed-loop algorithm, known as adaptive pressure control (see Chapter 2). This targeting scheme is available in most modern critical care ventilators under multiple names (e.g., PRVC, autoFlow, VC+, APV). A common confusion is that this is a volume-control mode, when, by the equation of motion, what is being controlled is pressure during a breath. A caveat with this targeting scheme is that in the presence of the patient's inspiratory efforts, the tidal volume may be higher than set, and the support provided by the ventilator may be inappropriately low.[50,51]
4. By controlling flow, the volume delivered is indirectly controlled. Because flow and volume are inverse functions of time (i.e., volume is the integral of flow and flow is the derivative of volume), controlling one controls the other. In simple ventilators, there is no feedback signal for flow, just a known flow for an adjustable amount of inspiratory time. On more sophisticated ventilators, the operator can regulate the shape of the inspiratory flow waveform. A square waveform will create higher peak airway pressures and will require less time to deliver the set volume (which may result in lower mean airway pressures) than a descending ramp pattern.[52–54] Some ventilators offer one waveform (e.g., the Dräger Evita XL offers only the

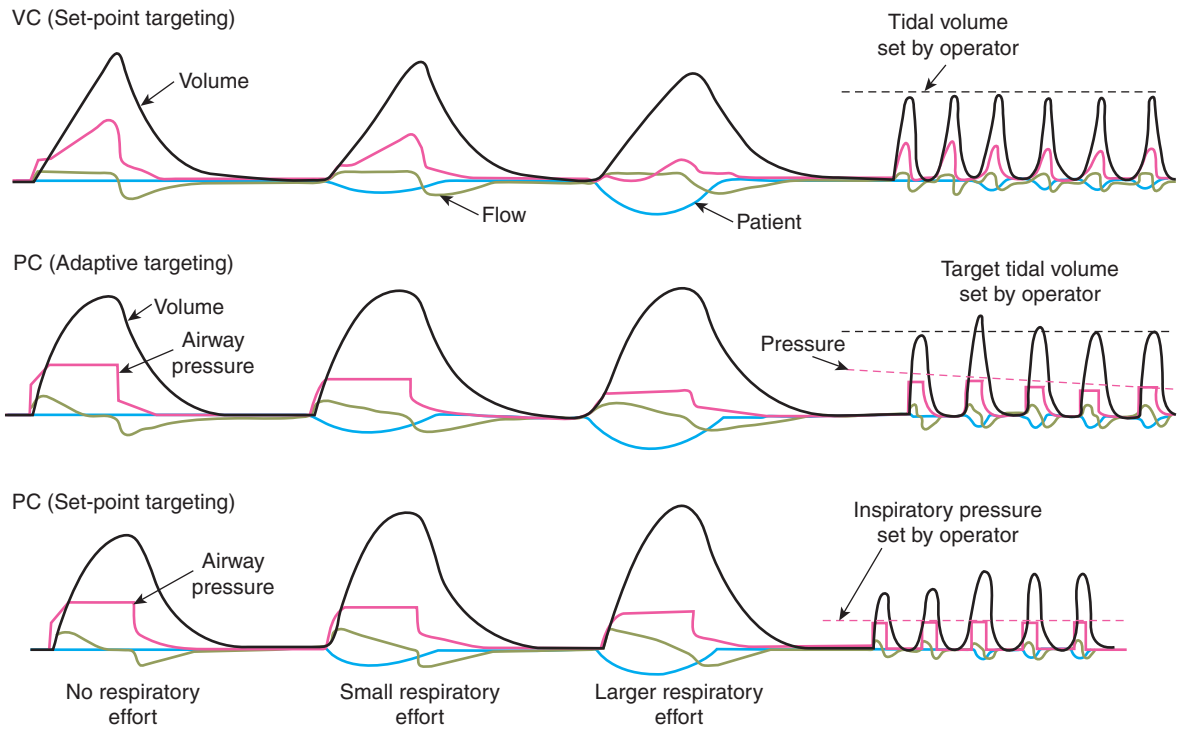


FIGURE 3-15 Volume delivery in volume control (VC) and pressure control (PC) modes using set-point targeting versus pressure control using adaptive targeting. Notice how tidal volume (flow) remains constant in volume control with set-point targeting in the setting of increased patient effort. In adaptive pressure targeting, the inspiratory pressure is adjusted by an algorithm to keep the tidal volume at a target. The tidal volume, however, may be larger if the patient effort is large enough. In set-point pressure targeting, the pressure remains constant, and the tidal volume increases in response to patient effort.

square waveform) others have more (e.g., the Hamilton Veolar offers 50% or 100% descending ramps, sinusoidal, and square).[55] Most current ventilators only provide the square waveform or a descending ramp profile.

Figure 3-15 compares volume delivery between standard volume and pressure control modes versus modes using adaptive pressure control.

MINUTE VENTILATION

In volume-control modes, the minimum minute ventilation is set by entering the tidal volume and respiratory rate. This assures that the patient will receive a minimum amount of ventilatory support. Some modes provide the option to enter a target minute ventilation (as a percent of the calculated minute ventilation for a given ideal body weight, adaptive-support ventilation [ASV]; e.g., Hamilton G5), while others will calculate it from the entered tidal volume and respiratory rate (mandatory minute volume [MMV]; e.g., Dräger Evita XL). The concept of automatically adjusting the ventilator settings to maintain a constant minute volume was first described by Hewlett and Plat in 1977.[56] As implemented, for example, on the Dräger Evita XL ventilator, MMV is a form of volume control–intermittent mandatory ventilation. The operator presets the target minute ventilation by setting tidal volume and frequency. The ventilator then monitors the total minute ventilation as the sum of the minute ventilations generated by mandatory and spontaneous breaths. If the total minute ventilation is below the target value, the mandatory breath frequency will increase. As long, however, as the spontaneous minute ventilation is at least equal to the target value, mandatory breaths will be suppressed. In this way, the proportion of the total minute ventilation generated by spontaneous breaths can range from 0% to 100%. As a result, MMV may be considered a mode of automatic weaning.

Another version of MMV was used on the Hamilton Veolar ventilator (now obsolete); the target minute ventilation was maintained by automatic adjustment of inspiratory pressure (adaptive pressure support). That mode was replaced by ASV on newer Hamilton ventilators.[49] ASV is the only commercially available mode to date that uses optimal targeting. It was first described by Tehrani in 1991.[57] The operator inputs the patient's height and percent of minute ventilation to be supported (25% to 350%). The ventilator then calculates the ideal body weight and estimates the required minute alveolar ventilation assuming a normal dead space fraction. Next, an optimum frequency is calculated based on work by Otis et al[9] that predicts a frequency resulting in the least mechanical work rate. The target tidal volume is calculated as minute ventilation divided by

TABLE 3-3: DETERMINANTS OF MINIMUM AND MAXIMUM MINUTE VENTILATION FOR SOME COMMON MODES

Mode Name	A/C	SIMV	MMV	ASV	Smart Care
Operation	Operator enters a set rate and tidal volume. Patient may trigger breaths above set rate.	Operator enters a set rate and tidal volume. Patient may breath in between mandatory breaths with or without assistance.	Operator enters a set rate and tidal volume. Patient may breath with or without assistance. If his minute ventilation falls below minimum, then mandatory breaths initiate at a set rate.	Adaptive pressure control breaths target tidal volume and rate according to mathematical model.	Pressure support is titrated based on expert rules to achieve the range $etPCO_2$.
Control variable	Volume	Volume	Volume	Pressure	Pressure
Breath sequence	CMV	IMV	IMV	IMV	CSV
Minimum minute ventilation	set $V_T \times$ set f	set $V_T \times$ set f	set $V_T \times$ set f	Targeted by ventilator based on operator-entered body weight.	Targeted by ventilator to maintain "comfort zone" based on V_T, f, and $etPCO_2$.
Maximum minute ventilation	Variable: $V_T \times$ total f	Variable: $V_T \times$ total f	Variable: $V_T \times$ total f	Variable but ventilator will reduce support if patient attempts to increase above estimated minute ventilation requirement.	Variable but ventilator will reduce support if patient attempts to increase above estimated minute ventilation requirement.

Abbreviations: *A/C*, assist/control; *ASV*, adaptive support ventilation; *$etPCO_2$*, end-tidal pressure of carbon dioxide; *f*, ventilatory frequency—total f reflects the sum of machine- and patient-triggered breaths; *MMV*, mandatory minute volume; *SIMV*, synchronized intermittent mandatory ventilation; V_T, tidal volume.

respiratory frequency (MV/f). In ASV, there are two breath patterns based on the patient's respiratory effort. If there is no patient effort, the ventilator delivers adaptive pressure-control ventilation; if there is patient effort, the patient receives adaptive pressure support. In both instances, the inspiratory pressure within a breath is controlled to achieve a target tidal volume.[49]

Table 3-3 summarizes the determinants of minimum and maximum minute ventilation for some common modes.

INSPIRATORY FLOW

The inspiratory flow can be adjusted by the operator on most ventilators that provide volume-control modes (see "Tidal Volume" above). In general, the ventilator operator will choose a peak flow and may have some waveform pattern options (e.g., square waveform or descending ramp). Although these settings appear simple, there are several points that may cause differences in performance and interpretation of data. First, the ventilator uses a microprocessor to control the delivery according to the preset tidal volume, inspiratory time, flow pattern, pressure limits, and ventilator-specific algorithms. During the breath, the flow delivery is adjusted according to a closed-loop feedback mechanism and proprietary software.[2] The consequence is a difference in performance among ventilator brands, even in the same mode.[27] Second, the interface may add confusion. For example, in the Dräger Evita XL, while on volume control, the operator will need to set the inspiratory flow, the inspiratory time, and tidal volume, whereas on the Hamilton G5, the options are customizable in three different ways! (Hopefully, all conducive to the same output.) The operator can enter (a) the *I:E* and the percent pause in inspiration, (b) the peak inspiratory flow and inspiratory time, or (c) the percent inspiratory time and plateau pause time. Underscoring that knowledge of the device used is essential. Finally, to add to the confusion, there are incorrect conclusions that sometimes permeate practice:

1. *In pressure-control mode, the flow is controlled as a descending ramp*. In a pressure-controlled breath, the volume and the flow are the manifestation of the respiratory system characteristics (resistance and compliance) and the patient's respiratory effort. If the patient is passive (no respiratory effort), the flow will decay exponentially (see Fig. 3-7, *A*). If the patient has a respiratory effort, the flow pattern will be variable, according to the characteristics of the patient effort, the ventilator settings (inspiratory pressure, pressurization algorithm, triggering, etc.), and the respiratory system characteristic. The only way to have a standard descending ramp is to select that waveform and have the computer control the flow delivery in volume control.

2. *The "autoflow" function adjusts the flow in a volume-controlled breath to the patient's demand.* Autoflow is available in Dräger Evita ventilators. It appears as an add-on for three modes of volume-control ventilation (controlled mechanical ventilation [CMV] or intermittent positive-pressure ventilation [IPPV], SIMV, and MMV). This "add on" is defined in the manual as automatic regulation of the inspiratory flow adjusted to the changes in lung conditions and to the spontaneous breathing demands.[58,59] What this "add on" does is turn the mode from a volume-control mode to an adaptive pressure-control mode. This is the same as being on PRVC on the Maquet ventilators. They all automatically adjust the inspiratory pressure to achieve a target tidal volume and because this is a pressure-controlled breath, the flow will be variable (see "Tidal Volume" above).

The inspiratory flow setting has importance at different levels. The work of breathing is related to the peak flow and the pressurization rate. The balance between patient and ventilator work of breathing will be affected by the inspiratory flow setting. In regards to cycling, high flows can lead to high peak inspiratory pressures (peak inspiratory pressure [PIP] is directly proportional to resistance, the higher the flow, the higher the PIP), which may lead to reaching the pressure or flow-cycling threshold and ending the breath prematurely.[59] But a more practical issue is this: does the flow-wave shape itself have any effect on patient outcome? Like most other questions about ventilator settings affecting patient outcome, after more than 30 years of research on this particular subject we still do not know the answer.

Studies from the early 1960s to early 1980s produced conflicting results, prompting Al-Saady and Bennett to design a better-controlled study, keeping tidal volume, minute ventilation, and *I:E* ratio constant.[60] They discovered that compared to a constant inspiratory flow, a descending ramp flow (what they and many subsequent authors have called "decelerating flow") resulted in a lower peak airway pressure, total respiratory resistance, work of inspiration, dead space-to-tidal volume ratio, and alveolar–arterial oxygen tension gradient. They also noted an increase in compliance and partial pressure of arterial oxygen (Pa_{O_2}) with no changes in partial pressure of arterial carbon dioxide (Pa_{CO_2}) or any hemodynamic variables. In 1991, Rau et al compared peak and mean airway pressure for seven different inspiratory flow waveforms (including square, ascending and descending ramps, and sinusoidal) under three different lung model conditions.[54] For all models, the descending ramp flow waveform produced the lowest peak and the highest mean airway pressures, whereas the ascending ramp produced the opposite: the highest peak and lowest mean values. When compliance was low, mean airway pressure increased as peak airway pressure increased. When resistance was high, peak airway pressure was more affected by the peak flow setting than the waveform setting.

In 1996, Davis et al[52] tested the hypothesis that a descending ramp flow waveform is responsible for improvements in gas exchange during pressure-control ventilation for acute lung injury. They compared volume control with a square or descending ramp waveform to pressure control with a square pressure waveform. Both pressure control and volume control with a ramp waveform provided better oxygenation at lower peak airway pressure and higher mean airway pressure compared to volume control with the square-flow waveform.

Polese et al[61] compared square, sinusoidal, and descending ramp flow waveforms in patients after open heart surgery. They found that Pa_{O_2} and Pa_{CO_2} were not affected by changes in waveform. Peak airway pressure was highest with the sinusoidal waveform while mean airway pressure and total work of breathing were least with the square waveform. Yang et al[53] applied square, sine, and descending ramp flow waveforms to patients with chronic obstructive pulmonary disease (COPD) and found that the descending ramp reduced inspiratory pressure, dead space-to-tidal volume ratio, and Pa_{CO_2}, but increased alveolar–arterial oxygen tension difference with no change in arterial oxygenation or hemodynamic variables.

Our own experience is that many clinicians prefer the descending ramp flow waveform when using volume control modes, with the observation that patients tend to be more comfortable, perhaps because of the higher flow earlier in the inspiratory phase.

Figure 3-16 illustrates an algorithm that can be used to adjust inspiratory flow to improve patient–ventilator synchrony.[62]

PERCENT SUPPORT

Proportional-assist ventilation (PAV)[63] delivers pressure-control breaths with a servo targeting scheme (see Chapter 2).[49] The pressure applied is a function of patient effort: the greater the inspiratory effort, the greater is the increase in applied pressure (Fig. 3-17). The form of PAV implemented on the Dräger Evita XL ventilator (called proportional pressure support) requires the operator to input desired assistance values for elastance and resistance. PAV implemented on the Puritan Bennett 840 ventilator (called PAV +) uses a different algorithm. It automatically calculates the resistance of the artificial airway, and combines resistance and elastance such that the operator enters only a single value representing the percentage work of breathing to be supported.[64] The design differences between proportional pressure support and PAV + lead to significant performance differences.[65]

NEURALLY ADJUSTED VENTILATORY SUPPORT LEVEL

NAVA is a mode that applies airway pressure proportionately to patient effort based on the voltage recorded from diaphragmatic activity. The "NAVA level" is the constant

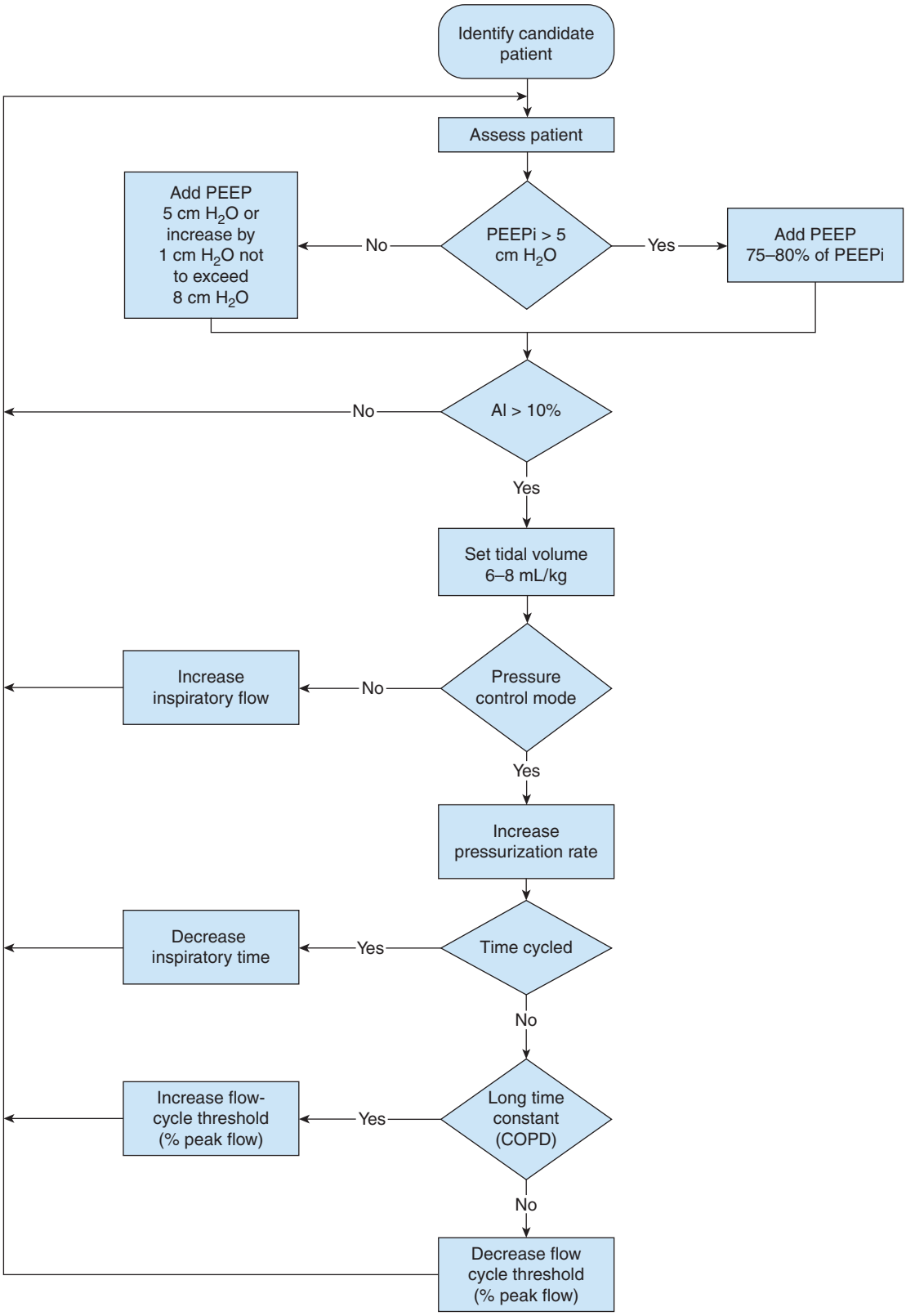


FIGURE 3-16 Algorithm for improving patient–ventilator synchrony. *AI*, asynchrony index, percent of inspiratory efforts that failed to trigger a breath; *COPD*, chronic obstructive pulmonary disease; *PEEPi*, intrinsic PEEP (aka auto-PEEP). (Modified from, with permission, Sassoon CSH. Triggering of the ventilator in patient-ventilator interactions. *Respir Care.* 2011;56(1):39–48.)

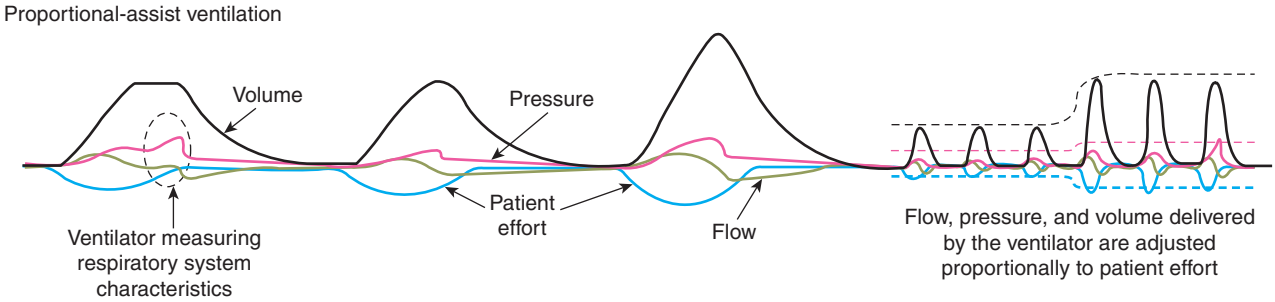


FIGURE 3-17 Pressure, volume, and flow waveforms for proportional assist ventilation.

of proportionality (gain) between voltage and airway pressure. The operator enters the NAVA level, then the ventilator delivers pressure equal to the product of gain and the Edi. In simple terms, it states how much pressure the patient will receive for each microvolt of diaphragmatic activity:

$$Paw(t) = Edi(t) \times \text{NAVA level} \quad (8)$$

where $Paw(t)$ is the airway pressure (cm H_2O) as a function of time (t), $Edi(t)$ is the electrical activity of the diaphragm as a function of time (t), in microvolts (μV), and the NAVA level is the operator-set level of support in cm H_2O/μV. The range is 0 to 30 cm H_2O/μV.

The NAVA level is set according to the operator ventilation goals, level of inspiratory pressure support, tidal volume, apparent patient work of breathing, or respiratory rate. Recently, Roze et al[66] proposed using the maximum Edi during a spontaneous breathing trial to help set the NAVA level (Fig. 3-18). By titrating the NAVA level to the a target Edi, the goal is to avoid excessive diaphragmatic unloading as well as respiratory muscle fatigue.

AUTOMATIC TUBE COMPENSATION

Automatic tube compensation (ATC) is a mode that compensates for the flow-dependent pressure drop across an endotracheal tube during inspiration and expiration. It is thus intended to reduce or eliminate the resistive work of breathing imposed by the artificial airway. ATC is an add-on feature on several ventilators. When ATC is activated, the ventilator supplies airway pressure in proportion to the square of flow times, a gain factor that is determined by the size of the endotracheal tube. Because flow is positive during inspiration and negative during expiration, ATC pressure either adds to inspiratory pressure or subtracts from expiratory pressure (Fig. 3-19). Some ventilators calculate and display tracheal pressure as airway pressure minus ATC pressure. ATC can be used alone or added to the ventilating pressure in pressure-control modes. Interestingly, the way ATC was implemented in the intensive care unit ventilators is different from the original concept, where negative pressure could be applied during exhalation.[67,68]

Cycle Variables

The inspiratory phase of a mechanical breath ends (cycles off) when a threshold value for a measured variable is reached. This variable is called the *cycle variable*, and it ends the inspiratory time. Cycling is characterized by the initiation of expiratory flow. The cycle variable may be preset (by the operator or the ventilator manufacturer), or automatically defined by the ventilator. Many different signals are used, for example, time, volume, pressure, flow, diaphragmatic signal, and thoracic impedance.

INSPIRATORY TIME

Inspiratory time is defined as the period from the start of inspiratory flow to the start of expiratory flow. Inspiratory time has two components; inspiratory flow time (period when inspiratory flow is above zero) and inspiratory pause time (period when flow is zero). In pressure-controlled or volume-controlled breaths, the inspiration is cycled (terminated) when the set inspiratory time elapses. In spontaneous modes of ventilation (NAVA, PAV, pressure support), the inspiratory time is dependent on the patient's own neurally determined inspiratory time, level of support, cycling rule (flow, pressure, time, diaphragm activity), and safety rules (maximum set inspiratory time).

Inspiratory time is usually an operator-entered input but some modes of ventilation can automatically set it and change it based on expert rules and closed-loop feedback algorithms. Two notable algorithms are ASV and Adaptive I-Time. In ASV (Hamilton G5), the inspiratory time is automatically set at one expiratory time constant (of the measured respiratory system characteristics and it is never shorter than 0.5 second or longer than 2 seconds). In the Adaptive Flow and Adaptive I-Time in the Versamed iVent (GE Healthcare, Madison, WI), the ventilator automatically adjusts the inspiratory time and inspiratory flow to maintain a target *I:E* ratio of 1:2 and deliver the operator-set tidal volume.[49]

In volume-control modes, there are four possibilities for setting inspiratory time:

1. Operator sets tidal volume and inspiratory flow: inspiratory time is equal to the tidal volume divided by mean inspiratory flow.

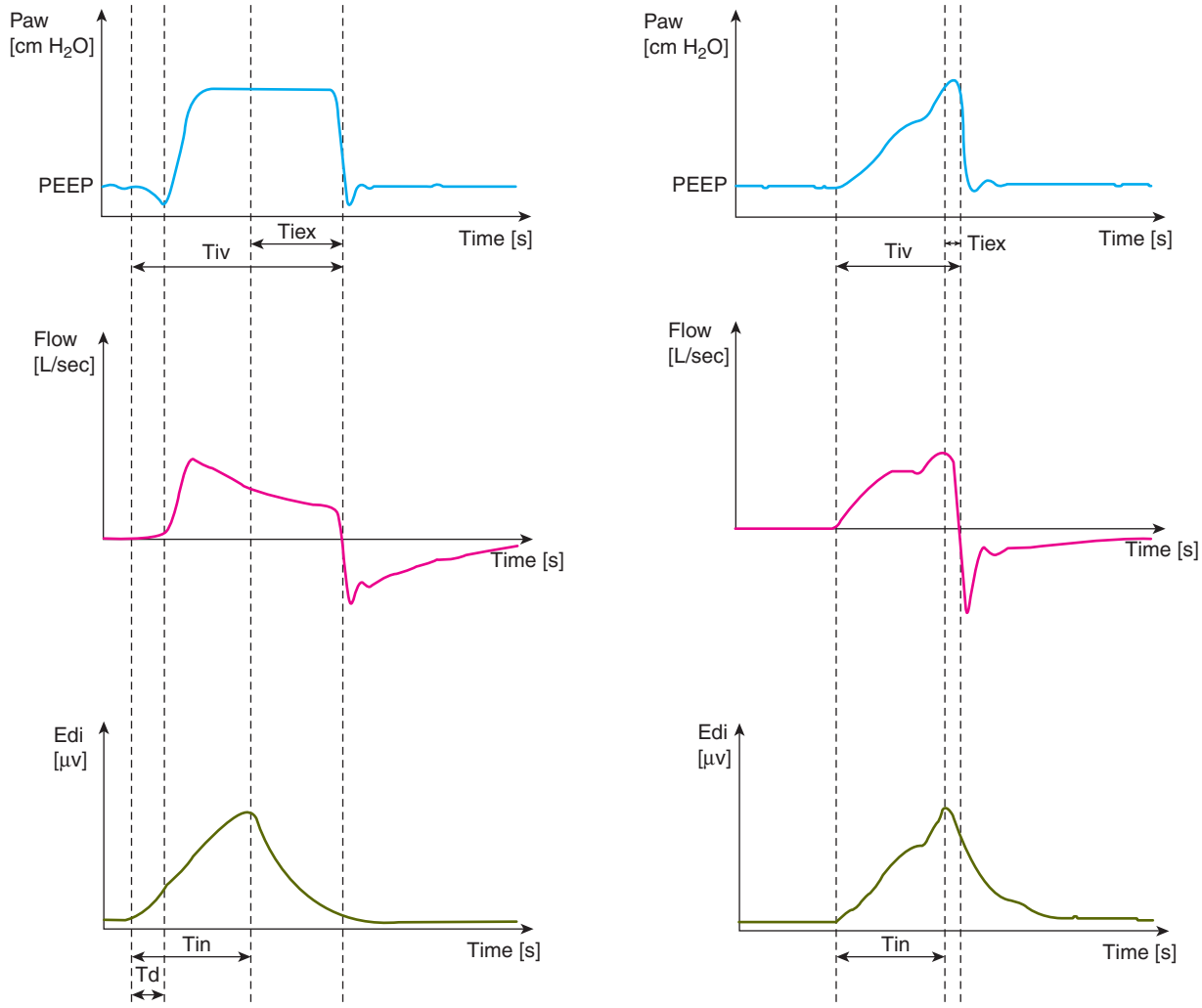


FIGURE 3-18 Airway pressure, flow, and electrical diaphragmatic activity curves in pressure support (left) and in neurally adjusted ventilatory assist (right). *Edi*, electrical activity of the diaphragm; *PEEP*, positive end-expiratory pressure, *Td*, trigger delay; *Tiex*, inspiratory time in excess; *Tin*, neural inspiratory time; *Tiv*, ventilator pressurization time. (Reproduced, with permission, from Piquilloud L, Vignaux L, Bialais E, et al. Neurally adjusted ventilatory assist improves patient–ventilator interaction. *Intensive Care Med.* 2011;37(2):263–271.)

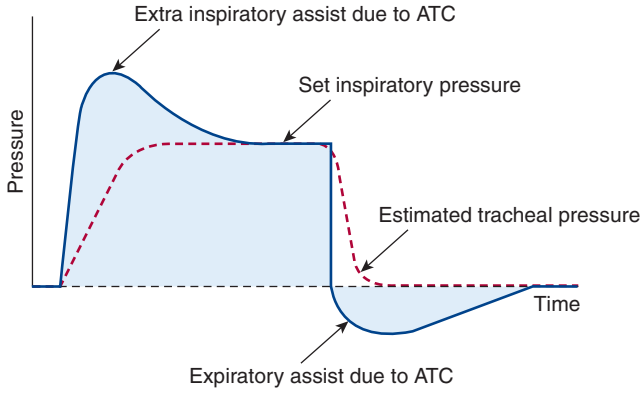


FIGURE 3-19 Pressure waveforms illustrating automatic tube compensation (ATC). (Modified, with permission, from Dräger Medical AG & Co. KG. *Infinity V500 Operator's Manual.* Luebeck, Germany.)

2. Operator sets tidal volume and inspiratory time: mean inspiratory flow is equal to the tidal volume divided by the inspiratory time.
3. Operator sets tidal volume, inspiratory flow, and inspiratory time: if the inspiratory time is longer than the inspiratory flow time (set tidal volume divided by set flow), then an inspiratory hold is created and the pause time is equal to the inspiratory time minus the inspiratory flow time. For example, if the tidal volume is 600 mL (0.6 L) and the set inspiratory flow is 60 L/min (1L/s) then the inspiratory flow time is (0.6/1 = 0.6 s). Now, if the operator also sets the inspiratory time to 1 s, an inspiratory pause is created and it lasts 1.0 − 0.6 = 0.4 s.
4. On some ventilators, the operator sets pause time directly.

In pressure-control modes, the operator presets the inspiratory time directly for mandatory breaths. Thus, prolonging the inspiratory time causes the ventilator to decrease the expiratory time, possibly resulting in air trapping, larger tidal volumes, or cycle asynchrony. One must remember that the effect on tidal volume of the inspiratory time in a pressure-control breath will depend on the respiratory system characteristics (i.e., the time constant). Thus, a patient with a long time constant (high compliance and/or high resistance) will require a longer inspiratory time to achieve full pressure equilibration, cessation of flow, and complete tidal volume delivery.

Figure 3-16 illustrates an algorithm that can be used to adjust inspiratory time to improve patient–ventilator synchrony.[62]

Inspiratory Pause. The inspiratory pause is the period during which flow ceases but expiration has not begun (see inspiratory time). The expiratory valves are closed during this period. The inspiratory pause time is part of the inspiratory time. It is also named plateau time (PB 840, Covidien, Mansfield MA), Pause time (Servo i, Maquet,) or Pause (G5, Hamilton Medical). When set directly, pause time may be entered in seconds or as a percentage of the inspiratory time. When it is activated, most ventilators will display a plateau pressure (i.e., static inspiratory hold pressure). Increasing the inspiratory pause time will increase the mean airway pressure and thus the time the lung is exposed to volume and pressure. This may have a positive effect on oxygenation and ventilation by increasing mixing time and decreasing dead space.[69,70]

I:E Ratio and Duty Cycle. *I:E* is the ratio of inspiratory time to expiratory time (Fig. 3-20).

$$I:E = T_I : T_E = \frac{T_I}{T_E} \tag{9}$$

The *I:E* can also be described as the duty cycle or percent inspiration. In engineering, the duty cycle is defined as the time spent in active state as a fraction of the total time. In mechanical ventilation, the active state is the inspiratory time, and the total time is the sum of the inspiratory and expiratory times. It is expressed as a percentage. The larger the percentage, the longer the inspiratory time in relation to the total cycle time.

$$Duty\ Cycle = \frac{T_I}{T_I + T_E} \times 100 \tag{10}$$

One can convert one to the other by the following formula:

$$I:E = \frac{Duty\ Cycle}{100 - Duty\ Cycle} \tag{11}$$

Example: A duty cycle of 50% is an *I:E* of 1:1, a duty cycle of 33% is an *I:E* 1:2.

The relevance of *I:E* is highlighted in the context of the time constant. The time constant is a measure of how quickly the respiratory system can passively fill or empty in

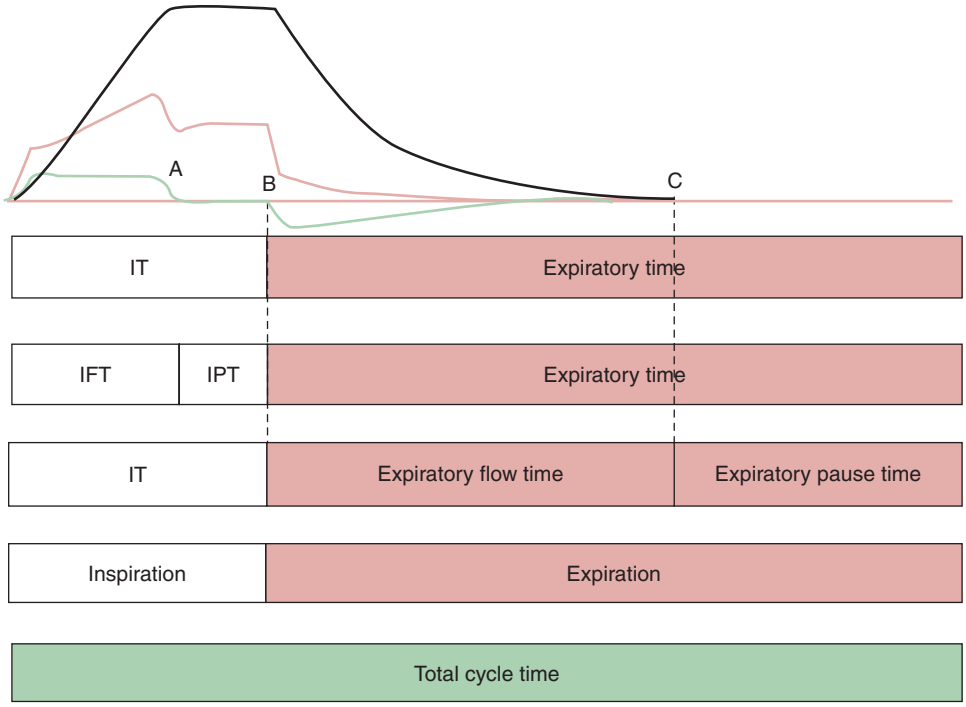


FIGURE 3-20 Divisions of the inspiratory and expiratory periods. A volume-controlled breath is depicted. **A.** End of inspiratory flow. **B.** Start of expiratory flow. **C.** End of expiratory flow. *IFT*, Inspiratory flow time; *IPT*, inspiratory pause time; *IT*, inspiratory time.

TABLE 3-4: EFFECT OF LUNG CONDITION ON TIME CONSTANT AND EXPIRED VOLUME

	Expiratory Time (s)			Expiration		
Time Constant	Normal Lung	ARDS	COPD	Tidal Volume Remaining (mL)	Tidal Volume Exhaled	Tidal Volume Remaining
0	0	0	0	500	0	100
1	0.780	0.510	1.000	184	63%[a]	37%
2	1.560	1.020	2.000	68	86%	14%
3	2.340	1.530	3.000	25	95%	5%
4	3.120	2.040	4.000	9	98%	2%
5	3.900	2.550	5.000	3	99%	1%

Abbreviations: *ARDS*, acute respiratory distress syndrome; *COPD*, chronic obstructive pulmonary disease.
[a]The exact value is $(1 - e^{-1}) \times 100\%$.

response to a step change in transrespiratory pressure.[23] It is calculated as the product of resistance and compliance. The value obtained is the time that takes to achieve 63% of steady state. This percent change remains a constant, regardless of the combination of resistance and compliance. It follows that *each* time constant will lead to a 63% decrease or increase in volume. In Table 3-4, one can see the difference among time constants for different lung conditions. In COPD, the time constant is longer so the time required for exhalation is longer than for patients with acute respiratory distress syndrome. This table demonstrates the effect of the time constant during passive exhalation using previously published[71] expiratory time constants for three conditions (normal lung was 0.78 seconds, for acute respiratory distress syndrome 0.51 seconds, and for COPD 1 second). In this example, expiration starts from a lung volume of 500 mL above functional resting capacity. When expiratory time equals one time constant, 63% of the tidal volume will be exhaled, leaving 37% of the tidal volume yet to be exhaled.

The *I:E* ratio can be an operator-entered value, or just displayed as a calculated value based on common scenarios for mandatory breaths:

- Preset *I:E* ratio and frequency.
- Preset inspiration time (T_I in seconds) and frequency (breaths/min). The frequency sets the ventilatory period ($1/f$) and the expiratory time is the period minus T_I:

$$I:E = T_I : [(60 \div rate) - T_I] \quad (12)$$

- Expiratory time and inspiratory time are fixed:

$$I:E = T_I : T_E \quad (13)$$

Note: some ventilators will synchronize inspiration and/or expiration of a mandatory breath if the patient effort is detected in a trigger/cycle window (e.g., SIMV or APRV), which may alter the *I:E* from the expected value based on settings.

PRESSURE

Pressure cycling occurs when the ventilator reaches a preset peak airway pressure. Pressure cycling is most often a safety feature (i.e., an alarm setting) with current modes of ventilation. When a preset high-pressure alarm threshold is crossed, the ventilator will cycle the ventilator. The goal is to prevent the patient from exposure to hazardous pressures. Pressure cycling without an alarm is the normal operational state for some devices (e.g., VORTRAN automatic resuscitator).

VOLUME

Volume cycling occurs when a preset volume is reached. This occurs when the operator sets a tidal volume in volume-control modes. Volume cycling implies that inspired volume is monitored by the ventilator's control system during inspiration and compared to a threshold value (the set tidal volume). But on some ventilators, despite the setting of a tidal volume, the actual cycle variable is time, that is, the time it takes to deliver the set tidal volume with the set inspiratory flow. Manufacturers seldom make this distinction clear in the operator's manual.

Volume cycling can also be found as a default safety feature. In PAV + (Covidien PB 840 ventilator), one of the cycling criteria is volume. Once the operator-preset high inspired tidal volume limit is reached, the ventilator cycles the breath and alarms.

FLOW

Flow cycling occurs when a preset flow or percentage of the peak flow is reached for pressure-control breaths. Flow cycling is most commonly found with the pressure-support mode but can be added as an "advanced setting" in other pressure-control modes on at least one ventilator (Avea, CareFusion). The flow-cycling threshold preset by the operator has been given many names: expiratory trigger sensitivity (Hamilton ventilators); trigger window (Engstrom Ohmeda); inspiratory termination peak inspiratory flow (Dräger Evita XL); expiratory threshold (Newport); flow termination (Pulmonetics LTV ventilators); PSV cycle (Avea, CareFusion); inspiratory cycle off (Servo *i*, Maquet); Ecycle (V200 respironics); and E sens (PB 840, Puritan Bennett).

During a breath in the pressure-support mode, the ventilator provides enough initial flow to achieve the set inspiratory pressure. The initial flow is high and then decays exponentially. Some ventilators have a preset default value for flow cycling (range: 5% to 30% of peak inspiratory flow); others allow the operator to adjust it (range: 1% to 80% of peak inspiratory flow). Only one device (e500, Newport Medical, Costa Mesa, CA) has automatic adjustment of the flow-cycling criteria. This device has a proprietary algorithm called FlexCycle. It will change the cycle criterion from 10% to 50% of peak flow based on measurements of airway pressure, the expiratory time constant, and expert-based rules applied through a closed-loop system.[72]

A default cycle criterion of 25% to 30% of the peak flow seems inappropriate as a "fit all" measure. The goal of adjusting the flow-cycling criterion is to avoid expiratory asynchrony.[59] In expiratory asynchrony, the ventilator ends inspiration before or after the patient inspiratory effort. We must remember that flow is a manifestation of the respiratory system characteristics, respiratory muscle effort (inspiratory and expiratory) and the integrity of the lung-ventilator circuit. If the respiratory system has a prolonged time constant, a standard flow-termination criterion may be inappropriate as it will prolong inspiration. That may be the case for patients with COPD, where the standard criterion of 25% may be too low, and lead to expiratory asynchrony and increased work of breathing.[73,74] Finally, a leak in the ventilator circuit (mask) or in the patient (endotracheal cuff or a bronchopleural fistula) may lead to lack of decay in the flow curve and thus asynchrony.[72]

Figure 3-16 illustrates an algorithm that can be used to adjust the flow-cycle threshold to improve patient–ventilator synchrony.[62]

DIAPHRAGMATIC SIGNAL

One goal of mechanical ventilation is to improve the patient–ventilator synchrony. In a perfect setting, the beginning and end of an assisted breath would be correlated with the neural signal driving the inspiratory muscles. In conventional ventilation that is rarely the case.[75] NAVA attempts to achieve this goal with the use of an electromyogram signal obtained from the diaphragm (Edi). As diaphragmatic activity decreases, so does the amplitude of the Edi curve. When it decreases below 70% of the peak signal (or 40% when the peak value is low), inspiration is cycled off. As a safety feature there is also a time-cycling mechanism. Piquilloud et al compared NAVA versus pressure support with the usual cycling criteria and found a significant improvement in expiratory synchrony (see Fig. 3-18).[76]

Baseline Variables

The baseline variables are the variables controlled during the expiratory time. Expiratory time is the period from the beginning of expiratory flow to the initiation of inspiratory flow. Flow and volume are not directly controlled during this period on any current ventilator. The most common value controlled is pressure relative to atmospheric pressure (zero-gauge pressure).

Positive End-Expiratory Pressure

The PEEP is established by the ventilator exhalation valve. A common source of confusion is the term continuous positive airway pressure versus PEEP. Continuous positive airway pressure is generally considered to be a mode on mechanical ventilators (or a mode of treatment for sleep apnea), whereas PEEP is the elevation of the baseline pressure during any mode of ventilation and is generally a setting for a mode. Until recently, the selection of PEEP has been a relatively arbitrary process and the meaning of "optimum PEEP" is debatable.[77] Now, Hamilton Medical has developed the INTELLiVENT system for the G5 ventilator that uses an algorithm for automatic targeting of PEEP and FIo_2. A closed-loop algorithm based on expert rules defines the response of the ventilator to measured ventilation variables, end-tidal carbon dioxide and pulse oximetry.

P-Low. P-low is one of the settings entered for so-called "bilevel" modes like APRV (Fig. 3-21). P-low is just another name for PEEP. Similar to PEEP, the settings are dependent on the user. There is, however, a large discrepancy with the objective of PEEP. In APRV, P-low is set to zero.[78] The goal is to maintain lung recruitment with the use of auto-PEEP induced by short T-low settings. P-low can also be set based on the biphasic model,[79] where complete exhalation is allowed and P-low is then set with the same goals as PEEP.

Expiratory Time. Expiratory time is defined as the period from the start of expiratory flow to the start of inspiratory flow. As stated above, the expiratory time is commonly dependent on the set inspiratory time, and set respiratory rate. It is rarely a fixed value. This occurs because making it a fixed value would produce, in most modes, changes in the inspiratory phase (inspiratory time, flow, and pressure). The most common exception to this is on ventilators that offer some form of APRV/biphasic pressure-control mode where expiratory time is set as "T-low."

T-Low. With exception of APRV/biphasic, in all the modern modes of ventilation the expiratory time is dependent on the inspiratory time and frequency; it is not an operator-set value. In APRV/biphasic, the operator sets the time spent at lower pressure, that is, exhalation (see Fig. 3-21). T-low can be set by the operator based on the peak expiratory flow,[78] targeting exhaled tidal volume or allowing complete exhalation.[80] Setting T-low sets the time trigger threshold for mandatory breaths. Among the methods described in setting T-low in APRV, targeting percent of peak expiratory flow (%PEF) is perhaps the most promoted method. The goal is to set the T-low short enough to avoid full exhalation,

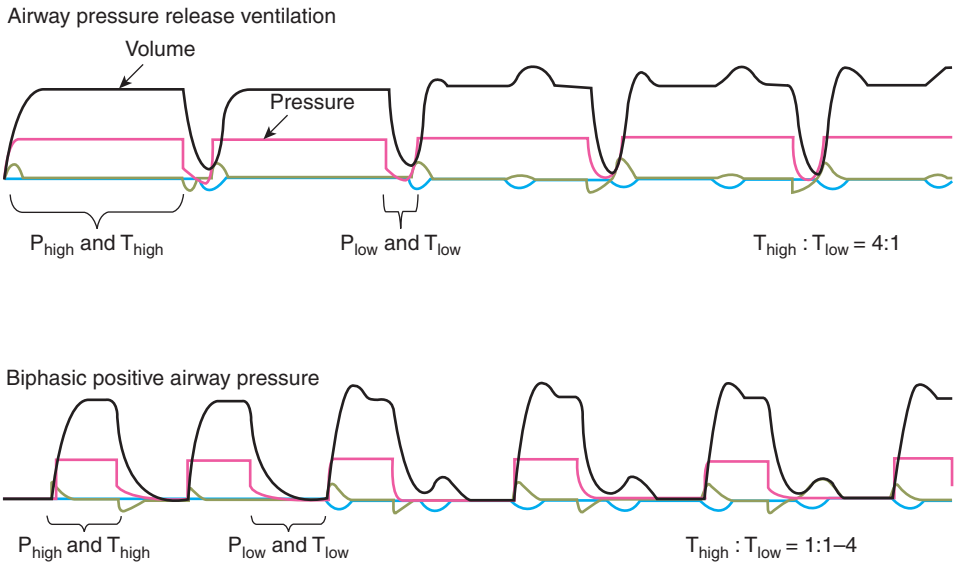


FIGURE 3-21 Differences in P-low and T-low settings for airway pressure release ventilation and biphasic positive airway pressure. Notice the difference in I:E ratio. The operator enters P-high, P-low, T-high, and T-low. The patient may breath spontaneously. Green curves show flow and blue curves show inspiratory effort.

thereby generating air trapping.[29] Adjusting T-low on the ventilator to manually maintain %PEF at 50% to 75% may be a tedious process, which may seem simple on paper, but in a spontaneously breathing patient can become a true challenge. Newer ventilators, like the Dräger Evita Infinity V500, have attempted to make the process easier by allowing the operator to set a trigger threshold based on a percentage of peak *expiratory* flow.

Alarms

Ventilator alarms bring unsafe events to the attention of the clinician. Events are conditions that require clinician awareness or intervention. Events can be classified according to their level of priority.[81]

Immediately life-threatening events are classified as Level 1. They include conditions like insufficient or excessive gas delivery to the patient, exhalation valve failure, control circuit failure, or loss of power. Level 1 alarm indicators should be mandatory (cannot be turned off by the operator), redundant, and noncanceling.

Level 2 events range from mild irregularities in machine function to dangerous situations that could threaten patient safety if left unattended. Some examples are failure of the air–oxygen blending system, inadequate or excessive PEEP, autotriggering, circuit leak, circuit occlusion, inappropriate *I:E* ratio, and failure of the humidification system. Alarms in this category may be self-canceling (i.e., automatically turned off if the event ceases) and are not necessarily redundant.

Level 3 events indicate changes in the amount of ventilator support provided to the patient consequent to changes in the patient's ventilatory drive or respiratory system mechanics and the presence of auto-PEEP. These events often trigger the same alarms as Levels 1 and 2.

Level 4 events are based entirely on patient condition. They may include events such as changes in gas exchange, dead space, oxygenation, and cardiovascular functions. Ventilators generally monitor these events and external monitors are required for alarms (the exception being exhaled carbon dioxide-level alarms built into the ventilator display).

Currently, ventilators do not display alarm settings as levels of priority. Instead, they tend to lump them all together on one screen that shows alarm limits and controls for changing them (Fig. 3-22). How to set alarm thresholds is a complicated topic that has been studied but for which little information is available regarding mechanical ventilation. The goal is to minimize false alarms and maximize true alarms. A high false alarm rate leads to clinician habituation and can also lead to inappropriate responses. In a recent study of an intensive care unit, 1214 alarms occurred and 2344 tasks were performed. On average, alarms occurred six times per hour; 23% were effective, 36% were ineffective, and 41% were ignored.[82] In another intensive care unit study, alarms occurred at a rate of six per hour. Approximately 40% of the alarms did not correctly describe the patient condition and were classified as technically false; 68% of those were caused by manipulation. Shockingly, only 885 (15%) of all alarms were considered clinically relevant.[83] Although these studies did not address mechanical ventilator alarms specifically, it is not hard to imagine similar results for such a study.

Ventilator alarms are usually set by the operator as either an arbitrary absolute value or a percentage of the current value. Examples would be airway-pressure alarms (high and low) set at the current value plus or minus 5 cm H_2O or low and tidal volume/minute ventilation set at plus or minus 25% of the current value.[81] The problem is that the parameters for which alarms are important, and these three

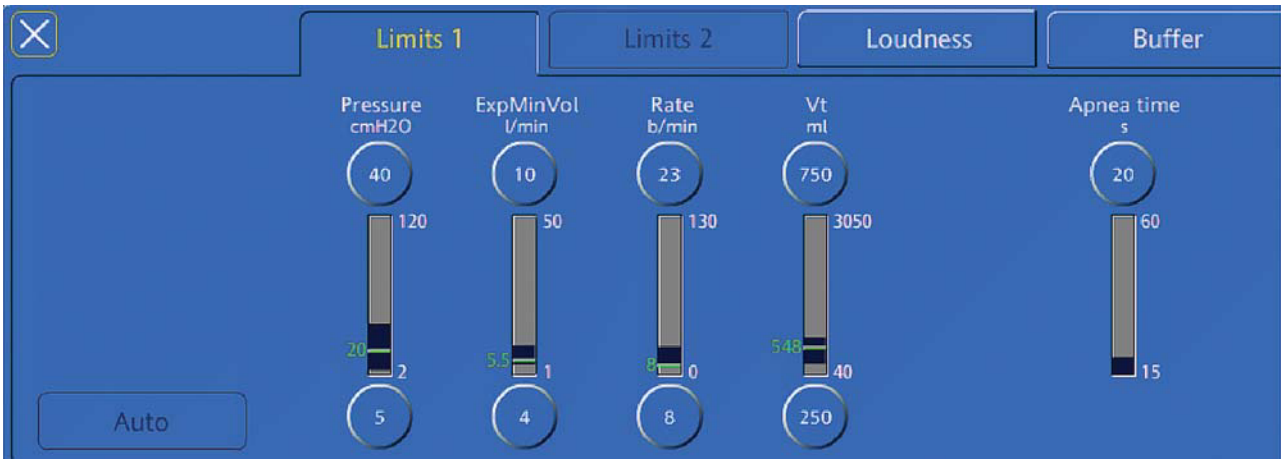


FIGURE 3-22 Alarm screen from the G5 ventilator. (Reproduced, with permission, from Hamilton Medical.)

in particular, are highly variable, with significant portions at extreme values.[84] Thus, limits set as absolute values or percentages may reduce safety for some extreme values while increasing nuisance events for other values. An alternative approach might be a type of "smart alarm," whereby the alarm limits are automatically referenced to the current value of the parameter such that extreme values have tighter limits. Further research is needed to identify optimization algorithms (i.e., minimize both harmful and nuisance events).

VENTILATOR OUTPUTS (DISPLAYS)

Display Types

Ventilator output displays represent the values of monitored parameters that result from the operator settings. There are four basic ways to present the monitored data: as numbers, as waveforms, as trend lines, and in the form of abstract graphic symbols.

NUMERIC VALUES

Data are most commonly represented as numeric values such as FIo_2, peak, plateau, mean and baseline airway pressures, inhaled/exhaled tidal volume, minute ventilation, and frequency. Depending on the ventilator, a wide range of calculated parameters may also be displayed including resistance, compliance, time constant, airway occlusion pressure at 0.1 second (P0.1), percent leak, *I:E* ratio, and peak inspiratory/expiratory flow (Fig. 3-23).

TRENDS

Many ventilators provide trend graphs of just about any parameter they measure or calculate. These graphs show how the monitored parameters change over long periods of time, so that, for example, significant events or gradual changes in patient condition can be identified (Fig. 3-24). In addition, ventilators often provide an alarm log, documenting such things as the date, time, alarm type, urgency level, and events associated with alarms, for example, when activated and when canceled. Such a log could be invaluable in the event of a ventilator failure leading to a legal investigation.

WAVEFORMS AND LOOPS

Many ventilators display waveforms (sometimes called "scalars") of airway pressure, volume, and flow as functions of time. Such displays are useful for identifying the effects of changes in settings or mechanics on the level of ventilation.[85] They are also very useful for identifying sources of patient–ventilator asynchrony, such as missed triggers, flow asynchrony, and delayed or premature cycling.[86] They can also display one variable against another as an *x-y* or "loop" display. The most common loop displays show pressure on the horizontal axis and volume on the vertical axis, or volume on the horizontal axis and flow on the vertical axis. Pressure-volume loop displays are useful for identifying optimum PEEP levels (quasistatic loops only) and over distension. Flow-volume loops are useful for identifying the response to bronchodilators. Figure 3-25 is an example of a composite display showing numeric values, waveforms, and loops.

ADVANCED GRAPHICS

As ventilators have become more complex, their displays have become more confusing and difficult to use. A recent trend is to move away from the traditional display screens in favor of a more integrative approach using creative graphic elements. For example, one study showed that observers detected and treated obstructed endotracheal tubes and auto-PEEP problems faster with graphical rather than conventional displays. They also reported significantly lower subjective workloads using the graphical display.[87]

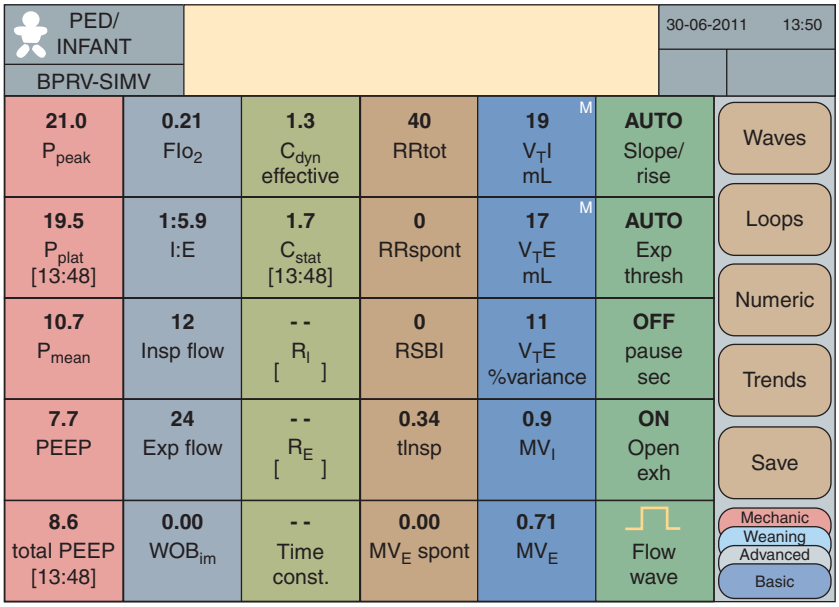


FIGURE 3-23 Digital display of monitored and calculated parameters from the Newport e360 ventilator. (Reproduced, with permission, from Newport Medical.)

Hamilton Medical was the first to make use of innovative picture graphics on their G5 ventilator. They created a graphic representation of the lungs, called a "dynamic lung panel," that visually displays information about resistance and compliance by the shape and color of the lungs and airways (Fig. 3-26). This panel is supplemented by a unique graphic, called the "vent status panel," which displays key parameters (e.g., oxygenation, ventilation, and spontaneous breathing activity). Furthermore, the display shows when each item is in or out of an acceptable zone and for how long. This makes weaning status easy to identify. Preliminary data[88] suggest that this display reduces the time required for clinicians to identify common problems, for example, normal, restrictive, and obstructive lungs; occluded endotracheal tube, right main-stem intubation, pre-spontaneous breathing trial (SBT), SBT in progress, and post-SBT phase.

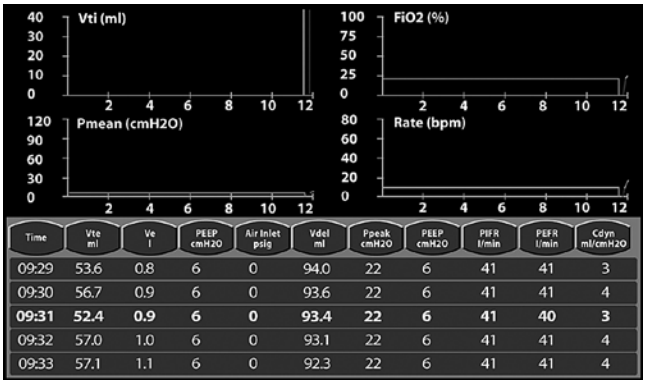


FIGURE 3-24 Trend display from the Avea ventilator. (Reproduced, with permission, from CareFusion.)

Dräger Medical recently introduced a similar graphic display called "Smart Pulmonary View." The shapes of the graphic elements quickly indicate relative values of respiratory system resistance and compliance as well as the balance between mandatory and spontaneous breaths (Fig. 3-27). Digital values are also displayed.

THE FUTURE

Better Operator Interfaces

As modes have become more complex, the operator interfaces on ventilators with computerlike displays has become cumbersome. Multiple options for control settings tend to get lost in layers of different screen views. Worse, screen views are often customizable such that if strict control is not exerted by an individual hospital department, each ventilator will be "stylized" by individual operators and chaos will ensue. Clearly, flexibility is a double-edged sword.

Very few studies have been published on ease of use or the problems with current displays. We need to identify optimal ways for ventilator displays to provide three basic functions: to allow input of control and alarm parameters, to monitor the ventilator's status, and to monitor the ventilator–patient interaction status. There is a long way to go before the user interface provides an ideal experience with these functions. This may be a fruitful area of future research.[6,8]

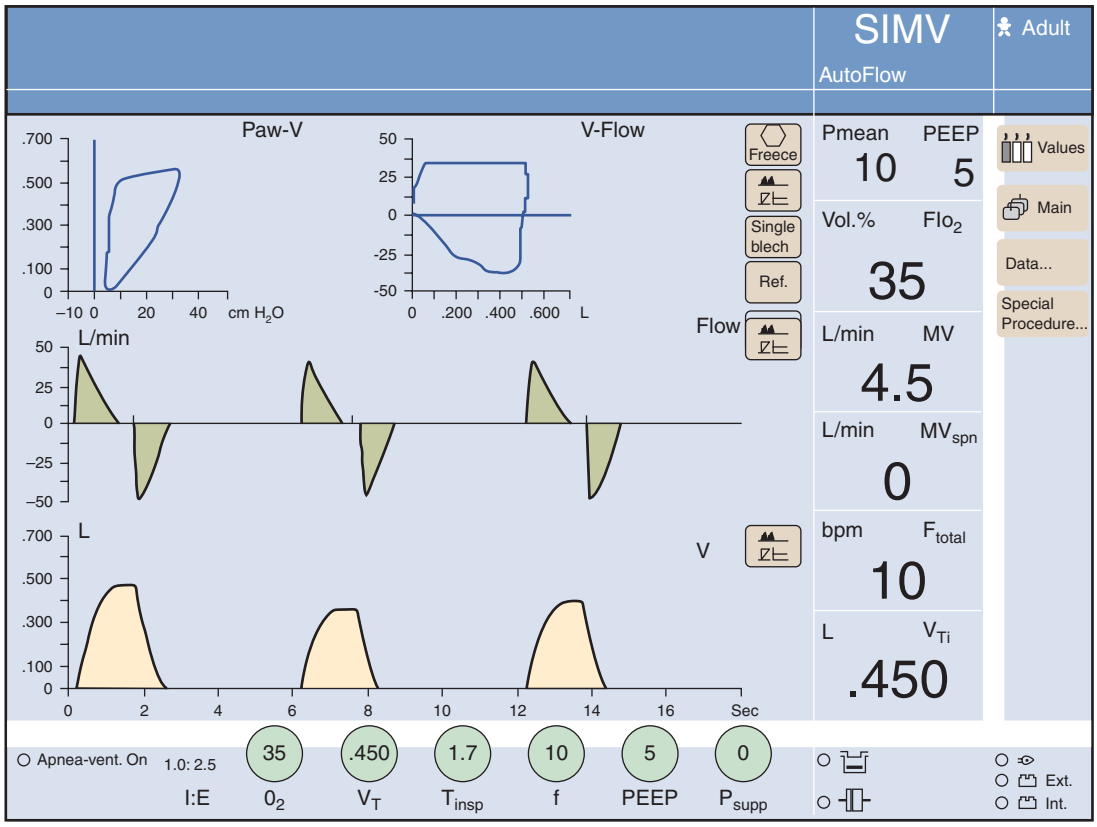


FIGURE 3-25 An example of both scalar and loop displays. (Reproduced with permission from Draeger Medical GmbH, Luebeck, Germany.)

Better Patient Interfaces

The interface between a modern ventilator and the patient is a piece of plastic tubing, that is, the "patient circuit," whose design has not changed much in several decades.

FIGURE 3-26 Example of picture graphic display from the Hamilton G5 ventilator showing the dynamic lung panel and the vent status panel. (Reproduced, with permission, from Hamilton Medical.)

Certainly, humidification systems using heated wires and automatic-temperature control have evolved, but we still are not capable of measuring and directly controlling a primary variable of gas conditioning: humidity. Indeed, after all this effort at evolving humidification systems, there are data to show that simple, unheated circuits provide better humidification of inspired gas.[89] In addition, the compliance of the patient circuit degrades the accuracy of flow delivery and must be "compensated" for by complex mathematical algorithms. It seems to us that a major revolution in patient-interface design would be to simply make the patient circuit a permanent part of the ventilator and treat water molecules the way we treat molecules of oxygen, nitrogen, helium, and nitric oxide. But to do this, ventilator manufacturers would have to merge with humidifier manufacturers and collaborate in systems design rather than seeing the patient circuit and humidifier as devices separate from the ventilator (see Chapter 2).

Better Targeting Systems

Chapter 2 provides a conceptual framework and suggestions for better targeting systems of the future. In essence, evolution in this area involves more and better sensors and the software algorithms required to manage the data they provide. The clear trend here, both in basic research and

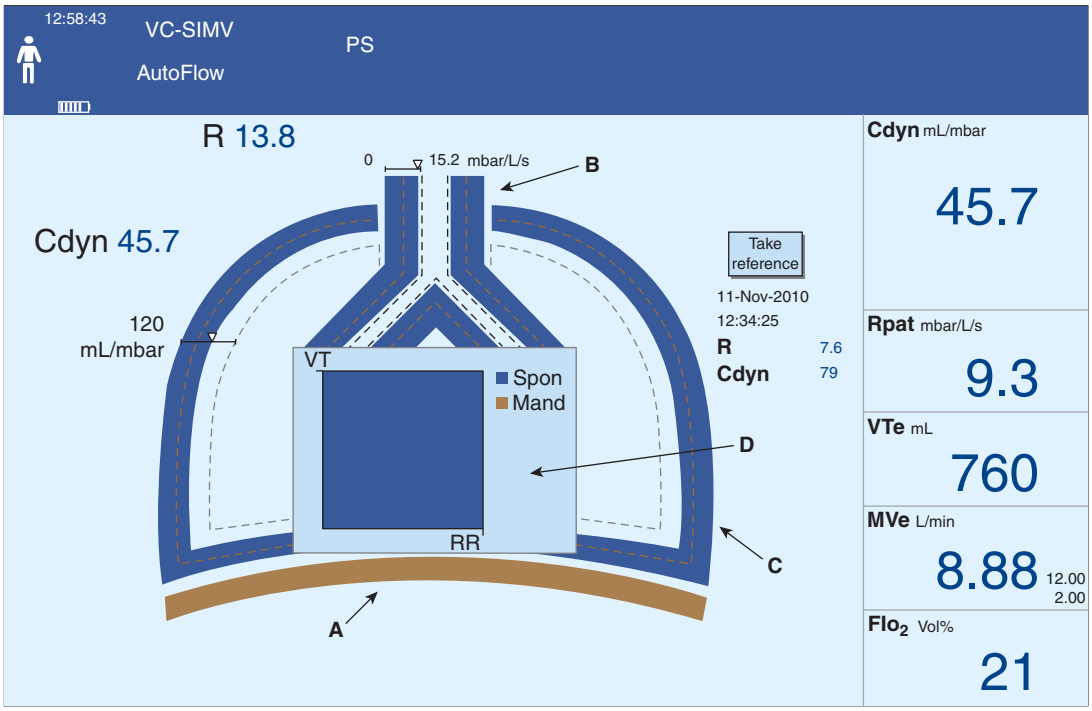


FIGURE 3-27 Example of picture graphic display from the Dräger Evita Infinity V500 ventilator showing the Smart Pulmonary View. **A.** The movement of the diaphragm indicates synchronized mandatory breaths or supported (triggered) breaths. **B.** The blue line around the trachea indicates the resistance R. The higher the resistance, the thicker the line. The numeric value is also displayed. **C.** The blue line around the lungs indicates the compliance Cdyn. The higher the compliance, the thinner the line. The numeric value is also displayed. **D.** Diagram displaying the relationship between spontaneous breathing and mandatory ventilation. The following parameters are displayed in different colors: spontaneous tidal volume (*VT spon*), spontaneous respiratory rate (*RR spon*), mandatory tidal volume (*VT mand*), and mandatory respiratory rate (*RR mand*). (Reproduced, with permission, from Draeger Medical GmbH, Luebeck, Germany.)

commercial applications, is to develop "closed-loop" targeting systems based on mathematical models of physiologic processes, or artificial intelligence, or combinations thereof, with the goal of automating the moment-to-moment adjustment of ventilator output to patient needs. The best example so far is a mode called INTELLiVENT-ASV (G5 ventilator, Hamilton Medical) and is currently available only in Europe.

This mode is an improvement on the optimal targeting scheme that is the basis of the mode called ASV (see Chapter 2). Like ASV, INTELLiVENT-ASV is a form of pressure control intermittent mandatory ventilation using adaptive-pressure targeting to automatically adjust inspiratory pressure to maintain a target tidal volume, which, in turn, is selected by an optimization model. An "optimal" targeting scheme attempts to either maximize or minimize some performance metric.[49] In the case of ASV, the ventilator attempts to select a tidal volume and frequency (for passive ventilation) that minimizes the work rate of ventilation for the patient's particular state of lung mechanics. As the lung mechanics change, the ventilatory pattern changes. ASV requires that the operator input the patient's weight, however, so that the ventilator can calculate an estimated minute ventilation requirement. The operator must also manage PEEP and FIo_2. INTELLiVENT-ASV takes ASV a step further by adding input data from end-tidal CO_2 monitoring and pulse oximetry. These extra data, along with advanced targeting software algorithms, allow the ventilator to automatically select and adjust minute ventilation, PEEP, and FIo_2. This makes INTELLiVENT "… the world's first complete closed-loop ventilation solution that offers automated adjustment of oxygenation and ventilation."[90]

Along with the new targeting systems, this mode also provides a unique operator interface that Hamilton refers to as the "Ventilation Cockpit," an apparent reference to the "autopilot" feature in airplanes. The interface is designed to facilitate understanding complex information in a visually intuitive way. In addition to displaying the usual digital parameters and waveforms, the new mode offers several other screens. The "Dynamic Lung" screen integrates data on lung mechanics, end-tidal carbon dioxide ($P_{ET}CO_2$), and pulse oximetry (SpO_2), and offers a metric called the "heart–lung interaction" index (Fig. 3-28). A graphic element called the "Ventilation Map" plots $P_{ET}CO_2$ against peak airway pressure as shown in Figure 3-29. Another display, the "Oxygenation Map," is very similar to the Ventilation Map: it provides detailed information about the oxygenation status based on the major physiologic input, as measured by pulse oximetry (SpO_2), and the resulting treatment ($PEEP/FIo_2$).

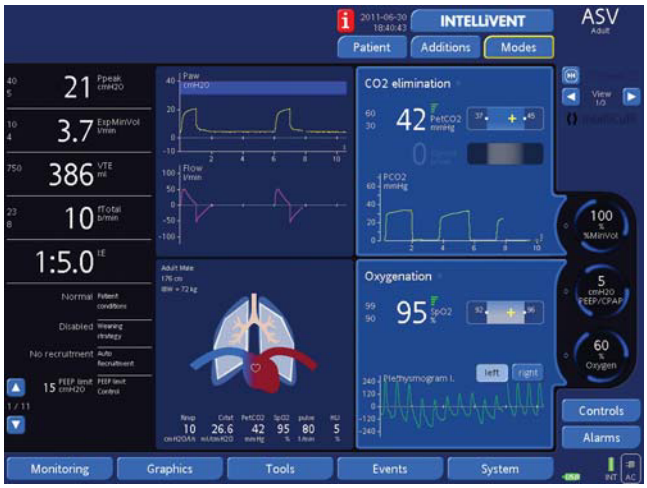


FIGURE 3-28 The operator interface of the Hamilton G5 ventilator with INTELLiVENT-ASV option, called the "Ventilation Cockpit." This screenshot shows the "Dynamic Lung" display including the "Heart–Lung Interaction" index. (Reproduced, with permission, from Chatburn RL. Computer control of mechanical ventilation. *Respir Care.* 2004; 49:507–515.)

Early studies of INTELLiVENT show that compared to ASV, patients ventilated with INTELLiVENT spent more time with optimal ventilation and less time with nonsecure ventilation. In addition, INTELLiVENT delivered lower volumes and pressures for equivalent gas exchange.[91] In another preliminary study, patients managed with INTELLiVENT spent less time with nonsecure and nonoptimal ventilation (3%) compared to conventional ventilation (47%, $P = 0.03$) after cardiac surgery.[92]

We speculate that modes of the future will continue this trend toward automation and include protocols for automatic weaning for various populations of patients. They will provide means for communication with electronic medical records and move us closer to integration of vast amounts of data into useful information for measurable improvements in patient outcomes. This process, however, will present significant challenges to vendors and end users to develop standardized vocabularies, taxonomies, and data transfer protocols in order to assure higher levels of accuracy, security, and usability.

FIGURE 3-29 The operator interface of the Hamilton G5 ventilator with INTELLiVENT-ASV showing details of ventilation and oxygenation management.

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